# GAIT ANALYSIS OF NORMAL AND DIFFERENTLY ABLED SUBJECTS FOR REHABILITATION

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**BY**

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## CERTIFICATE

This is to certify that the thesis entitled **"Gait analysis of normal and differently abled subjects for rehabilitation"** is a record of the bonafide work done by PROTIMA NOMO SUDRO (212BM1476) which is submitted for partial fulfilment of the requirements for the degree of Master of Technology (MTech) in Biomedical Engineering at National Institute of Technology, Rourkela. To the best of my knowledge, the matter embodied in the thesis has not been submitted to any other University/Institute for the award of any Degree or Diploma.

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# ABSTRACT

This study investigates the human foot function for gait analysis and rehabilitation studies. It involves demonstration of dynamics during different gait activities. The biomechanical response of the investigation implies that the centre of pressure moves accordingly with the motion of foot such as heel strike and take-off. The respective ground reaction forces are measured using a force platform. The data from the force platform are collected using a data acquisition system and processed using Bioware™ software. Curves of acceleration-time, displacement–time, power and impulse are calculated from the ground reaction force record obtained from the force plate. The centre of pressure, friction, torque and moment during walking was also calculated. In this study 57 participants have participated including male, female and differently abled subjects. The participants are instructed to walk at different rates, perform vertical jumps over the force platform and walk with barefoot and in shod. The study also computes the different phases of gait cycle and forefoot function index to observe the impulse acting in the forefoot. The performance of the participants in vertical jumps are characterized by the flight height which is actually the difference between the height of the centre of mass at the peak of the jump and the height of the centre of mass at the time of take-off. From the investigation it is observed that the level of activation and force of the participant's leg muscle bearing the body weight determines the dynamic characteristic of human locomotion. The evolution of the walk and vertical jump was traced carefully and discussed for male, female and disabled persons, identifying the gait phases for each case.

**Keywords:** gait analysis, rehabilitation, kinematics, dynamics, shod.

# CHAPTER 1

# INTRODUCTION

## 1.1Background

Gait assessment provides quantitative contribution in clinical research. Basically gait assessment is done in order to distinguish the type of impairments, suggesting diagnosis based on the analysis, monitor the austerity of an injury or a disease and determine most appropriate treatment [1]. Gait analysis includes assessments of individuals ranging from neuromuscular disorders to high-level athletes [2]. To perform gait analysis; a multiaxial force platform is used. The force platform is a metal plate consisting of 4 force transducers which employs piezoelectric sensors. The piezoelectric sensors are capable of generating charge proportional to the load applied on it. The Force plate is a rectangular plate with 4 pedestals on 4 corners. Each pedestal holds a single force transducer. The force transducers give an electrical output proportional to the force exerted on the metal plate. Since, the output from the force platform is digitized. The force plate with piezoelectric sensors is suitable for dynamic force measurement and it is large enough to hold both the legs at a time. The force platform is capable of measuring the ground reaction forces exerted by the ground in response to the load applied in it. It is a six-component force platform which measures vertical forces and horizontal forces. For studying the biomechanical relationship between various gait activities like standing, walking, running, jumping, squatting, sprinting, etc force platform is used. The force-time record obtained using force platform is useful for demonstrating both qualitative and quantitative relations between displacements, velocity and acceleration.

## 1.2 Biomechanics of human movement

Biomechanics of human movement deals with the function of musculoskeletal system in correspondence to the application of mechanics in it. In order to accomplish movement/motion forces are needed. Biomechanics describes the internal and external forces acting on the human body and how these forces affect the stability of human movement [3]. Today biomechanics has evolved as a very important branch of science by unveiling the kinematics and kinetics involve in human movement. Biomechanics explains the control of human movement monitors the pathological improvements, enhances the improvement of an athlete. Locomotion is one of the basic necessities of life. The study of locomotion pattern of living beings is termed as gait analysis. Generally, all human beings perform gait by a process known as Bipedalism. The purpose of bipedalism is to provide mechanism for locomotion and maintain stability. Bipedalism is unstable, until a control system (postural control system) acts continuously on both the legs. And the balance control is maintained by visual, vestibular and proprioceptive systems: central nervous system and musculoskeletal system [4].The central nervous system consisting of brain, spinal cord and nerves controls each and every aspects of movement (specifically locomotion). It determines the performance of muscles, bones and joints. The visual system is not use only to see clearly but it also acquires some information which in turn generates field-of-view information for movement (locomotion) [5]. The vestibular system along with visual system helps in maintaining the body's state of motion. And in musculoskeletal system, the muscles work together along with the bones in order to support the weight and provide controlled as well as precise movement. Biomechanics deals with internal and external forces acting on the body and the effect of those forces in the movement of the body [6-7]. Human motion is studied by applying forward and inverse dynamics.  Biomechanical analysis is executed using one of the two analyses: kinematics and kinetics.

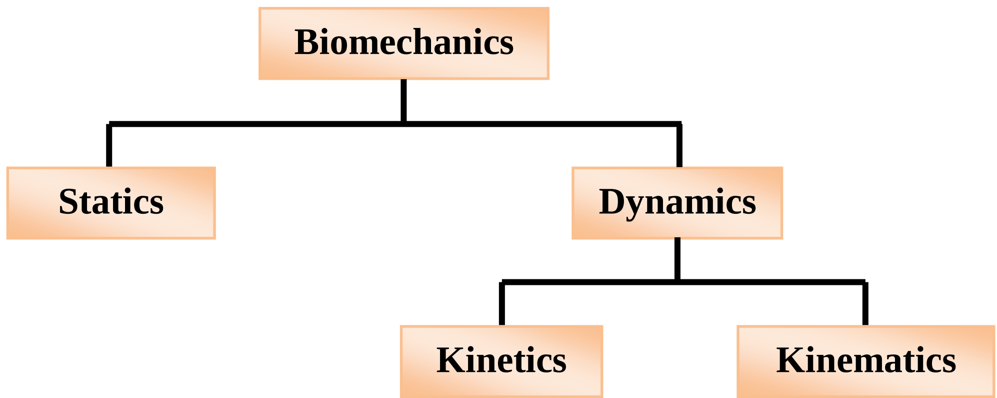


**Figure 1 Types of movement analysis**

In kinematics the cause of the movement is studied without considering the forces that cause the movement. Whereas in kinetics forces are involved in activities that are studied [8]. The classification of biomechanics for motion analysis is given in Figure 1.

## 1.3 Biomechanics of lower trunk

Besides the upper trunk, the biomechanics of lower trunk (hip-joint, knee-joint and ankle-joint) plays a very important role in human movement. The hip-joint also known as Coxa connects the lower trunk with the upper trunk. It is considered to be one of the strongest and largest joint in the human body. As it bears the whole load of upper extremity, the hip-joint must be stable enough to accomplish the functions in static and dynamic conditions. And the stability of the hip-joint is maintained by the multiaxial ball and socket joint existing between acetabulum and the head of the femur. The ball and socket joint has adapted in such a way that it meets the demands of gait activities like standing, walking, running, jumping, climbing staircase etc., with adequate range of motion. Ideally the range of motion consists of 120° flexion, 0° abduction and 20° lateral rotation [9]. Muscles along with flexors and extensors contribute to the weight bearing functions during static and dynamic conditions. The muscles around the hip-joint have distinct characteristics with large cross-sectional area for attachment resulting in high magnitude force. The main functions of hip-joint are adduction, abduction and rotation.

The knee-joint also provides significant contribution for locomotion. The knee-joint is a large and complicated joint which functions in erect position in absence of muscle contraction. The knee-joint lowers and elevates the body by rotating and flexing the bony structures along with the soft tissue muscles and ligaments providing stability.

The ankle joint is a hinge joint connecting the leg and the foot. It allows rotation only in one plane. The ankle joint consists of three important bones: tibia, fibula and talus. The ankle-joint participates in adduction, abduction, supination, pronation, dorsiflexion and plantar-flexion. The ankle-joint plays a crucial role in motion. It is the ankle-joint that transfers the force and energy required for forward progression. The ankle-joint acts as a base support by initially accepting the load applied (dorsiflexion) by the body followed by load transition (plantar-flexion) from one foot to the other.

## 1.4 Gait analysis

Gait analysis monitors every single parameters (velocity profiles, acceleration, centre of mass displacement, torque applied, moment of force, ground reaction force, centre of pressure, impulse, energy, power dissipation) relating to the motional activities. It examines the cause of performance parameters carefully by instructing the subjects to do many trials, sometimes it may last even for some days. From consistent performance, professionals check for any deviations in their gait, and then conclude whether it is a normal or abnormal gait.

Depending upon the continuous gait analysis (many trials) normal gait exhibits no deviations in their performance. It is found that their ground reaction force pattern shows two humps with smooth curve. However, abnormal gait shows variations from first observation of ground reaction force-time curve and it continues for all trials. Gait analysis deals with biomechanical problems resolved by using one of the two approaches: forward dynamics and inverse dynamics. In forward dynamics, the input signal is generated by the nervous system and based on the input signal the musculoskeletal system activates/reacts resulting in desired motional activities which is measure as the output. However in case of inverse dynamics, the data exhibited by a subject during motional activities (external forces) is considered as the input and by using the measured data further calculations like joint reaction forces, muscle moments etc are done. For solving inverse dynamic problems, the acquired data must be first analysed properly i.e. the data measured directly is termed as raw data according to Winter [10]. Raw data contains contaminated noise and hence, it should be treated by a process known as smoothening using digital filters. To prevent aliasing of the signal, data were collected at an adequate sampling rate. Adequate sampling rate is determined by Nyquist rate which states that “the sampling frequency must be greater than or equal to twice the maximum frequency component of the signal”. Amongst digital filters, Butterworth filter is more focused and widely used in biomechanics because of its flat frequency response (without any ripples) in the passband and roll off towards zero in the stopband.

Many people face problem in the lower extremity leading to an unusual gait pattern. Gait analysis is concerned with various activities like standing, walking, running, jumping, etc. Gait analysis is done in order to diagnose and monitor rehabilitation programs. Common gait related problems in normal healthy subjects occur due to injury, accident, etc. Some pathological gait is observed in differently-abled healthy subjects, like Spasticity, hemiplegic gait, etc. While analysing the biomechanics of gait: time, mass, force, moment and motion are very important variables because from these variables only motion analysis is done. Gait analysis is characterized by hip flexion/extension, knee flexion/extension and ankle flexion/extension.

Gait analysis is performed based on the standard cycle known as 'Gait cycle'. A gait cycle refers to the successive contact of the same limb during forward progression. Gait cycle consists of two different phases known as stance phase and swing phase shown in Figure 2. The stance phase also known as support phase, initializes at heel strike and persists until the toe is lifted off the ground. The stance phase in turn consists of three phases:

- Heel strike/ double support phase (both the foot are in partial contact with the ground- heel contact of the respective foot and toe-off of the alternate foot)
- Mid-stance (single support phase- the respective foot is completely in contact with the ground) and
- Toe-off/double support phase (metatarsal region and big toe of the foot is in contact with the ground).

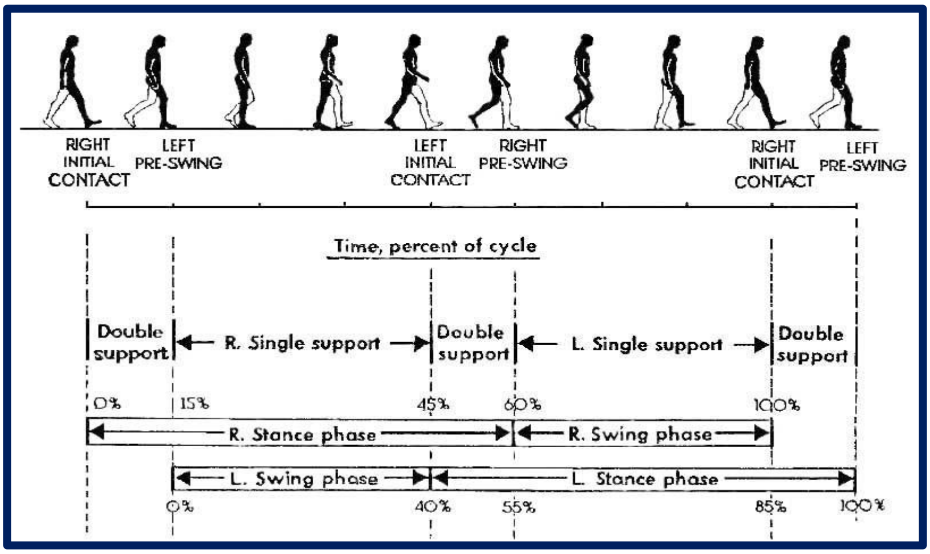


**Figure 2 Different phases of a gait cycle**

The swing phase also known as unsupported phase exhibits two important operations

- Acceleration and
- Deceleration.

Acceleration occurs when the swinging leg crosses the body followed by deceleration when the foot lands on the ground. Despite the stance phase and swing phase in gait cycle events like double support phase, single support phase, right leg stance phase, left leg stance phase, right leg swing phase, left leg swing phase, left initial contact and right initial contact also takes place. Double support phase is that event when both the legs/feet are in contact with the ground, whereas single support phase implies the event when only a single leg is in contact with the ground [11].

Most of the human movements depend upon the inter-relation with the ground. Whenever a human being performs motional activities like standing, walking, running and jumping, he/she continuously exerts the force of the body to the ground both vertically and horizontally. And in response of the body force the ground also exerts back an equal and opposite force to the body. This inter-relation is in agreement with Newton's 3rd law of motion. Due to this opposite and equal force exerted by the ground, a subject is able to maintain balance while doing any kind of motional activities. The force exerted by the ground in response to the body weight is known as ground reaction force. Since, it is the ground reaction force that stabilizes the motional activities, it is considered as the most important external force exerting on the body. By virtue, the ground reaction force is controllable by adequate co-ordination of the musculoskeletal system. So, it opens up a window for monitoring the pathological gait. Hence, the principle of Newton's law of motion is employed to understand the biomechanics of human movement. According to Newton's 2nd law of motion

$$\sum F = ma \qquad \text{..........} (1)$$

'F' here refers to summation of all the forces acting on the body (body weight and the ground reaction force) 'm' refers to the body mass and 'a' refers to the acceleration of the body's centre of gravity

$$\sum F = \sum F_{GRF} - W, \text{ W= mg}$$

'W' here refers to the body weight. Substituting above value in Eqn.1, the equation becomes

$$\sum F_{GRF} - mg = ma \qquad \text{..........} (2)$$

From equation (2), it is noticed that when the ground reaction force is equal to the body weight, the net force is neutralised leading to zero acceleration of body's centre of gravity. As the ground reaction force varies with the position of centre of gravity, the net force in turn becomes positive/zero/negative accordingly. When the centre of gravity moves downward the force plate measures a negative net force whereas when the centre of gravity moves upward it measures positive net force. The net force keeps on varying with position and direction of the movement.

## 1.5 Various motor activities

Standing is a posture during which the body is said to be perfectly balanced. This posture can be explained by using Newton's law of motion which states that an object remains in its state of motion until and unless it is acted upon by an external unbalanced force. As standing is a balanced state, the weight force vector acts in downward direction due to law of gravitation. So, while standing, the weight force vector acts to accelerate the subject in downward direction. But this acceleration does not occur because, according to Newton's 3rd law of motion to every action there is an equal and opposite reaction. Thus, while standing the ground/earth also exerts a force equal in magnitude and opposite to the direction of the weight force vector of the subject/person known as ground reaction force. Ground reaction force is very important gait parameter to validate pathological gait from normal gait [12]. Mathematically, during standing

$$W + F_g = 0 \qquad \text{...... (3)}$$

where W= weight of the person/subject (negative) $W = mg$, where m = mass of the subject and g is acceleration due to gravity (9.81m/s$^2$), $F_g$ (positive) represents the ground reaction force. Equation 3 implies that the vertical velocity of the person/subject is constant and is valid only when the person/subject is standing without any muscle activation . In case if the subject/person activates his/her leg extensor muscles while standing still, then the ground reaction force will vary resulting in greater or less ground reaction force compared to body weight and hence

$$W + F_g \neq 0 \qquad \text{....... (4)}$$

Next to breathing, walking is considered to be the most important function of the body. Walking is a motor function defined as the rhythmic pattern of alternating leg movements.

The main aim of walking is to move forward with desired speed employing least energy as possible. Here, the walking gait pattern can be best explained using a gait cycle. From a single gait cycle, the activities occurring during walking are determined [13]. Walking begins with heel strike and ends with toe lifting. Between these periods the activities like heel strike, flatfoot, midstance, heel rise and toe-off. Since walking is a conscious activity of the body it must be performed very easily and effortlessly [14]. At heel strike the heel pad compresses and acts as a shock absorber by propagating the load to the respective foot as soon as it lands on the ground [15]. This event is known as weight acceptance phase. At heel strike, the ankle remains in neutral position due to isometric dorsiflexors and the hip and knee are flexed to some degrees by eccentric knee extensors and isometric hip extensors. Immediately, after the heel strike the foot attains a flat position during which the ankle and knee are flexed to some degrees and the hip-joint is extended. During midstance phase the load of the body is completely transferred to the corresponding leg. This event is known as single support phase because the whole load of the body is carried by a single limb and the other limb is swinging. In midstance phase the ankle dorsiflexors are inactive and the plantarflexors are eccentrically contracting to maintain the rate of the limb for forward progression. At this stage the knee-joint is extended requiring very less contraction. And on the hip the concentric extensors and hip abductors acts to maintain the position of pelvis from descending in the frontal plane. Midstance phase is followed by heel off/rise during which the load is slowly transferred from the rear foot to the front/fore foot. At this point the ankle plantarflexors start to act concentrically, knee-joint starts to fortify for flexion and hip-joint is in a state of hyperextension.

In normal walking gait, stance phase holds for 60 percentage of the gait cycle and swing phase holds for 40 percentage of the gait cycle. However, during running the stance phase is found to hold for 40 percentage of the gait cycle and swing phase holds for 60 percentage of the gait cycle. Hence, the phases of gait cycle keep on varying with gait activities [16]. The gait cycle helps us in determining the rate of gait as well as distinguishes the pathological gait from normal gait. As human beings performed bipedal locomotion, with one foot on the ground and the other feet swinging during walking, balance becomes one of the challenging tasks. Depending upon the percentage of foot contact with the ground and the percentage of swinging, gait pattern is characterized as walking or running. Despite the gait pattern, the rate of gait activities can also be determined [17]. As walking is a rhythmic pattern showing twin-peak curve the net force, $\sum F$ varies with all the phases of leg motion.

Jumping is a gait in which the body of a subject is temporarily in air-borne phase. In jumping, two types are commonly studied: squat jumping and countermovement jumping. In countermovement jumping the subject initiates the jump by quietly standing in one position, makes a downward movement by crouching and then vigorously extends the leg-joints attaining a high position of centre of gravity. In countermovement jump $\sum F = 0$, then with crouching position $\sum F = negative$ and after extending the knees and ankle joints net force becomes $\sum F = positive$, when the force is very high compared to the body weight [9]. Compared to countermovement jump, squat jump is considered to be quite unusual because in this kind of jump the subject initiates the jump from crouch position and then extends its leg muscles. Generally, countermovement jump is performed most commonly. Squat jumping founds its application in skiing.

Running gait is characterised by its longer duration of swing phase compared to stance phase due to air-borne effect. In running the peak ground reaction force is observed to be 2 to 3 times higher than that of the body weight. In running gait, the force-time curve shows only a single peak instead of twin-peaked symmetric curve as observed in walking gait. The reason of a single peak curve in jumping is because of the force exerted during initial heel strike is attributed to the significant bending of the knees to absorb the impact force for lifting off the ground. In running gait, the leg does not stretch properly during the stance phase. With respect to that, the centre of mass reaches its maximum point, causing the ground reaction force to go down, resulting in a single peak ground reaction force curve. The phases of gait cycle are completely opposite to that of walking gait.

# CHAPTER 2

# LITERATURE REVIEW

## 2.1 Gait variability

The gait performed by human beings in daily life is not consistent through-out; sometimes a subject walking gait suddenly changes into running gait because of the adaptability of the body to invest least energy as possible. A subject's gait speed is dependent upon the leg length. An algebraic transformation based upon the walking pattern specifies that a subject cannot perform his/ her gait faster than $v = \sqrt{gl}$ [18]. The stability of human gait is characterized by balance and posture. Various balance tasks assessment techniques:

- State of balance during unperturbed standing
- State of balance during perturbed standing
- The expected and unexpected perturbations of centre of gravity
- State of balance during dynamic gait

and equipment (stop watch, video camera and force plates) are used to obtain quantitative information about the human gait [4]. Gait analysis is performed by observing the gait cycle and plays an important role in treatment decision- making clinically. A clinical test involves: video recording, physical exam, motion and muscle assessment, EMG tracing, kinetics and kinematics and measurement of metabolic expenditure. The integration of all the parameters and proper interpretation of the gait patterns by the experts categorize the normal gait from pathological gait. And based on the assessment, clinicians arrive at a specific treatment decision to cure a pathological gait [19]. Both internal and external forces are responsible to maintain the postural balance of the body. When no external force acts, the internal perturbations due to muscle contraction and other biological activities, body sway occurs sometimes leads to a forward fall [3]. Human gait is also stabilized by the type of shoes that a person wears. It has been reported that when a subject walks in barefoot, the subject acquires larger force as compared to that in shod. A subject walking in shod acquires lesser amount of force because of the shoe heel, which absorbs force and reduces the load on the foot. It is also reported that in barefoot walking the load on the forefoot is three times higher than in the hind foot [20]. Force plates are used to provide vertical ground reaction force component in order to realize the relationship between forces, acceleration, velocity and displacements. Force plates are tended to include more than 4 force transducers for improving the frequency response [21]. Biomechanical investigation of the back and front load carriage for exercise-related injuries are done to design human-centered load carrying systems. The load carriage system provides difference in the body centre of mass trajectory along the sagittal plane. The back load trajectory is found to be lower than the front load because of the larger forward lean angle with backload walking. Hence, the study provides an impact of the back loading and forward loading for improving the upright posture in gait [22]. The postural stability is assessed by instructing a subject to stand still on the force plate with eyes open one time and eyes closed another time. The variations define the use of eyes as one the important system to maintain postural stability [23]. The gait motion is highly modified by the activation of muscles along medio-lateral direction. Muscles correspond for more than 92% of the medial-lateral ground reaction force for all the walking gait speed. However, gravitational and other forces made very small contribution. Muscle correlates the acceleration along medio-lateral direction by exchanging the forces between medio-lateral ground reaction force cause by abductors and lateral ground reaction force cause by knee-extensors, plantarflexors and adductors. The muscles not only support the body weight and lead to forward progression but also regulate the medio-lateral acceleration of the body centre of mass during transition of the body weight from one foot to the other (double support phase to single support phase) [24]. Gait analysis focussed on the lower limbs as well as the upper body movement. The movement of the arm i.e swinging during walking is found to be providing stability of motion. It is quantitatively proved by using an index of the upper body movement while walking. The index is termed as SFD (shoulder fulcrum dispersion) and it is found that the at high speed gait SFD is higher and at low and normal self-selected speed gait the effect of SFD is very low. This characterizes the effect of hand swing in balancing the gait [25]. The ground reaction force measured by using force plate demonstrates the mechanics associated in gait. The centre of mass reveals a curve path during walking. The centripetal force measured provides a limitation to the speed of a subject’s gait. The centre of mass exhibits a simple spring during running showing a path similar to elastic bouncing ball [21]. Vertical jump is calculated and analysed by three methods:

- Flight time of the jump
- Applying impulse-momentum theorem to the force-time curve and

- Applying work-energy theorem to the force-displacement curve.

In vertical countermovement jump the more effective the downward phase the higher is the jump resulting in higher workdone in upward phase. This fact is highly implemented by volleyball and basketball players to achieve maximum height during vertical jump. Greater jump height is achieved when a subject performs a run-up for the jump which converts the kinetic energy into gravitational potential energy. Countermovement jump height is higher than squat jump[16]. The treatment scheme provided by M. L .Root [26] does not quantify the validity and reliability of the treatment. T.C.M. Poil and G. C. Hunt studied the reliability of measurement techniques, alignment of normal foot and the position of subtalar-joint during the midstance and toe-off phase during walking proposed by M.L. Root and use the tissue stress model and it is found to be very effective alternate for evaluation and treatment of foot disorders [27].

## 2.2 Design methods

A technique presents the design and use of two types of orthotic inserts: flat orthotic heel raise and contoured heel cup for determining the behaviour of heel pad during gait. The heel insert include ultrasonic transducer embedded in it for measuring the behaviour of the heel pad during gait. This technique gives an insight in designing and developing interventions for restoring the normal gait [28]. For the subjects with high medial-longitudinal arch, custom made insoles are found to be effective by reducing the forefoot and rearfoot pressures. The idiopathic pes cavus is the term generally use for high medial longitudinal arch [29].

## 2.3 Measurement approaches and Systems

The dynamic gait variables are measured by using instrumental metal plate known as force plate. Most of the biomechanics laboratory is equipped with Force platforms to measure the ground reaction forces involved in the motional activities. The force platform is three-component transducer consisting of piezoelectric sensors mounted on four corners of the plate. The principle of working is based on piezoelectric effect. The force plate is found to be capable of measuring ground reaction force, moment of force, torque, centre of pressure, friction and other parameters like velocity, acceleration, displacement etc [30].

## Objectives

The objective of the work is to

a) Study the gait parameters of normal and differently able subjects
b) Suggest remedial measures and to design an orthotics for the impaired subjects.

# CHAPTER 3

# EXPERIMENTAL METHODOLOGY

Prior to the experiment, the study was approved by the Institute Ethical Committee of National Institute of Technology, Rourkela. Informed consent is obtained from the subjects before acquiring their gait patterns.

## 3.1. Subjects

- For this study 57 subjects were consented. Among the 57 subjects 30 subjects are male and 24 subjects are female and 3 are differently able. Among 3, one subject is suffering from Spastic Quadriplegia
  - Neurological disease
  - Stiff and tight muscle
  - Joints are flexed to some degree
- Another subject is orthopedically handicapped (50%) due to an accident
  - Muscle grafted on both the legs.
  - Ankle-joint is plantarflexed with negligible dorsiflexion.
- And the third subject walks by leaning laterally to avoid the pain on one side due to an accident.

The patterns obtained from the subjects were used to analyze gait parameters in motor act. The subjects (abled and differently abled) were chosen for the study. The able subjects are free from any neurological and orthopaedic impairment, however differently abled subject exhibits pathological gait due to neurological disorders and other foot injury. The gait patterns like standing, jumping, normal (self-selected speed) walking, slow (slower than their normal speed) walking and fast (faster than their normal speed) walking gaits are recorded and analysed.

## 3.2 Force plate

Six-component multi-axial force platform (Kistler, Model No.9260AA6) used for gait analysis. The co-ordinate axes geometric configuration is shown in Figure 3. The force plate data is used in turn to calculate the joint reaction forces by inverse dynamics method. The force platform is a metal plate consists piezoelectric sensors on 4 corners. Each pedestal holds a single force transducer. Gait patterns are obtained at a sampling frequency of 1000 Hz. The subjects were properly instructed prior to the activities, and they were also asked to perform 2 to 3 trials before the acquisition of the data.

## 3.3 Data collection and analysis

The force data was recorded for walking. After 2 to 3 trials the natural gait pattern of the subjects was recorded. The subjects were made to walk through a 330 cm long path for duration of 15 seconds. The static and dynamic data were collected for the gait activities performed by the subjects.

## 3.4 Data analysis

The output voltage of the piezo element is given by $V = Q/C$, where C is the capacitance of the element and Q is the charge induced by a force applied in the plate. The force transducers give an electrical output proportional to the force exerted on the metal plate. Since, the output from the force platform is analog an additional interface generally Data Acquisition System (DAQ) is connected to the force platform to convert analog signal into digital signal [8].

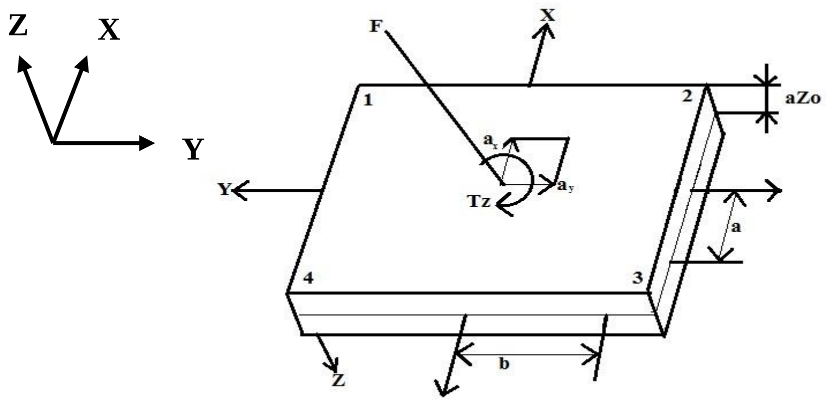


**Figure 3 Geometric configuration of a force plate**

The force plate with piezoelectric sensors is suitable for dynamic force measurement and it is large enough to hold both the legs at a time. Kistler coordinate system is used for X, Y and Z-planes, X-represents the medial-lateral axis of coordinates, Y-represents anterior-posterior axis of coordinates (direction of motion) and Z-represents the proximal-distal axis of coordinates. All the calculations were performed based on Newton's 2nd law of motion given as

$$F_Z(t) = m \, . a_Z(t) \qquad \text{…….. (5)}$$

The profile parameters are computed by using the equations derived from Eq. 5. The acceleration –time record is obtained using body mass information and static acceleration as described in Eq. 6

$$a_z(t) = \frac{F_Z(t)}{m} - a_{oZ} \qquad \text{…… (6)}$$

where $a_{oZ} =$ static acceleration, $a_z$ = acceleration along the vertical direction, m is body mass and $F_Z$ = vertical force. The moment of force about Z-axis is calculated based on the direction represented as

$$M_Z = b * (-f_x 12 + f_x 34) + a * (f_y 14 - f_y 23) \qquad \text{.…... (7)}$$

where $M_Z$ = Moment of force along Z-axis, $f_x$ 12 = force along x-axis measured by sensor 1 and 2, $f_x$ 34 = force along x-axis measured by sensor 3 and 4, a and b are respective offsets. From Eq. 7 the torque is expressed as

$$T_Z = M_Z - F_Y * ax + F_X ay \qquad \text{……. (8)}$$

where $T_Z$ = vertical torque, $F_X$ and $F_Y$ are the forces along x and y axis.

The centre of pressure is expressed as:

$$a_x = -\frac{M_y}{F_Z} \text{ and } a_y = \frac{M_x}{F_Z} \qquad \text{…….. (9)}$$

where $a_x$= centre of pressure along x direction, $a_y$ = centre of pressure along y-direction, $M_y$ = moment of force along Y-axis, $M_x$= moment of force along X-axis and $F_Z$ = force exerted vertically.

## 3.5. Data smoothening

As the data recorded from the force plate with the help of an interface (data acquisition system) is contaminated with some degree of noise. The data recorded is termed as raw data and it must be treated before performing gait analysis [9]. We begin gait analysis by collecting the data after some trials with adequate frequency in order to prevent aliasing of the signal (higher frequency signal pose to low frequency signal), where we use sampling frequency of 1000Hz. The sampling frequency is chosen based on sampling theorem – Nyquist theorem. As the raw data collected from the system is noise contaminated, it is treated to smoothen by using digital filter. Of the many digital filters, Butterworth filter is used for smoothening biomechanical data which attenuates the frequencies above the specified cut-off frequency and allows the frequencies below the cut-off to pass through the filter. The sharpness of Butterworth filter is characterized by its order of the filter. And thus we use order, n=2 and also padded zeroes at the end of each signal

# CHAPTER 4

## RESULTS AND DISCUSSION

### 4.1 Performance parameters of a normal and pathological gait

The force exerted by the ground in response to the force applied by a subject is referred to as ground reaction force and this concept is based on Newton's third law of motion stating that for every action there is an equal and opposite reaction. Ground reaction force is considered as one of the most important gait parameter in clinical gait analysis. Ground reaction force pattern is capable of distinguishing a pathological gait from that of a normal gait. Hence to quantify it further we use Newton's 2$^{nd}$ law of motion $F = ma$, and it is noticed that when the ground reaction force is equal to the body weight, the net force is neutralised leading to zero acceleration of body's centre of gravity.

### 4.1.1 Typical Ground reaction force curve of a healthy and pathological gait

With the help of Newton's laws of motion we observe the ground reaction force curve of a normal subject as shown in Figure3. As there are only 2 force platforms, the subjects are instructed to perform gait across the path of 330 cm for 15 seconds. The notation R and L means right and left foot. The first gait cycle does not begin at 0.1 sec because the subject initiates the gait a single step away from the force platform. Thus, there is a gap from zero to first gait cycle and again there is a gap between 1$^{st}$ and 2$^{nd}$ gait cycle because the subject makes a step length after it lands from the force plate and turns around to walk back on the force plate by making another step length to reach the force plate.

Since the gait analysis is instructed to perform for 15 seconds, the subjects are asked to walk continuously across the force platform. For normal gait it is observed that a subject is able to perform three gait cycles in the given time duration. The ground reaction force shown in Figure 4 (R and L determines the respective right and left foot force component) corresponds to the vertical force component and anterior- posterior force component. The vertical component shows greater force magnitude followed by anterior-force component.

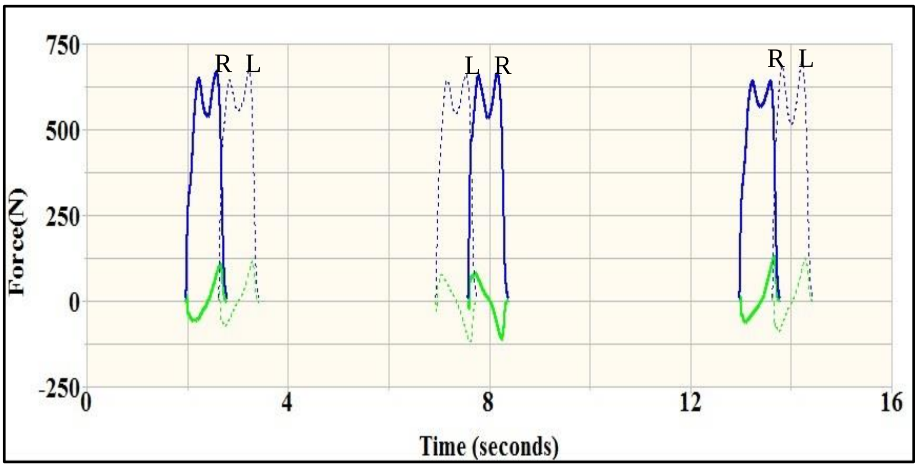


**Figure 4 Typical ground reaction force curve of a normal healthy gait**

When heel strikes the ground an impact force is exerted by the ground and during this time the entire body weight is impose on that planted limb. With heel strike the force curve increases gradually due to the acceleration of the planted foot carrying the entire weight of the body and the force magnitude during this time is around 40 % of the body weight. Heel strike is followed by midstance phase. Mid stance phase is that instant of gait phase when the foot covers maximum area by completely interfacing with the ground exhibiting maximum force. The midstance is followed by heel rise or toe-off phase when the entire body weight is transferred from the hind foot to the forefoot. During toe-off phase the foot overcomes the frictional force cause by the ground in order to accelerate the foot for forward progression. Thus the ground reaction force curve characterizes the motion of the foot. It initiates with heel contact, force magnitude increases gradually with the foot contact due to the transition of the load and shows a sharp peak when the foot is completely in contact with the ground. This is followed by a decrease in force magnitude due to heel rise making a plateau in the ground reaction force pattern. However the force magnitude starts to rise again because of the entire body weight acting on the forefoot in order to transit the body weight from the planted foot to the other. The toe-off phase is accompanied by swing phase of that limb, where the action of the swinging limb is compared to a projectile, neglecting the air-resistance. The characteristic nature of the ground reaction force curve obtained from a normal healthy subject walking at self-selected speed shows two symmetric humps signifying high stresses during mid-stance (ankle is in neutral state) phase and propulsive loading/ toe-off phase (ankle is in a state of plantarflexion). Since both the limbs exert same characteristics, only the single limb behaviour is discussed. Due to high rate of loading the impact force indication is not visible in the graph. Ground reaction force curve incorporates the three phases of walking i. e heel strike, midstance phase and toe-off phase. During heel strike only a portion of the foot surface is in contact with the ground. With initial heel contact the anterior-posterior force component (y-direction) undergoes negative phase and continues to be negative until the heel pad is raise to achieve momentum. The negativity of the anterior-posterior component is attributed to the braking motion of the leg by decelerating the motion of the swinging leg. At this stage the subtalar joint located between talus and calcaneous acting as a pedal puts a brake on the motion of the swinging leg with slight inversion. And then lift the hind foot putting pressure on the forefoot where the subject applies high impulse to overcome the frictional force by pushing the ground slightly backward. The other gait cycles also shows the same characteristic pattern. Other step lengths also follow the same procedure.

A pathological gait is shown in Figure 5 below whose gait is affected due to spastic quadriplegia. Spastic quadriplegia is a type of cerebral palsy and from many studies it is reported as well as observed that individuals with spastic quadriplegia acquaint upper and lower trunk muscle weakness and stiffness restricting their range of motion [11]. The weak and stiff muscles are influenced by the intrinsic mechanical properties of muscles and tendons and depending upon the degree of weakness, the capacity of muscle activation is reduced. The reduction of muscle activation in turn amplifies the antagonist muscle co-contraction. The anti-gravity muscle such as gastrocnemius muscle is found to be respectively shorter in length compare to a normal healthy subject. In many cases individuals with Spastic quadriplegia are not able to walk, however subject in this study is moderately functional in day to day life. From the knowledge of various studies, the ground reaction force-curve obtained from a differently abled subject characterizes a pathological gait pattern shown in Figure 5. From the ground reaction force curve, it is noted that the subject was able to complete only a single gait cycle in the given time duration, whereas a normal healthy subject was able to complete 3 gait cycles for the same time period. Here, the rate of loading of the planted foot is not significant and this elucidates reduced length of the anti-gravity muscle and Achilles tendon. Generally, tension is developed when muscles get stretched and muscles are stretched only when they are provided sufficient strain energy by the muscle fibres. The intrinsic muscular force restricts the stretching of the Gastrocnemius and hamstring muscles resulting in reduced force for attaining required motion [31]. As the joints are stiff to a certain degree the subject's force generated from the upper part (pelvic joint and knee joint) of lower extremity are not transferred entirely to the subject's ankle joint and as such the braking force in the anterior-posterior plane is negated for few milliseconds.

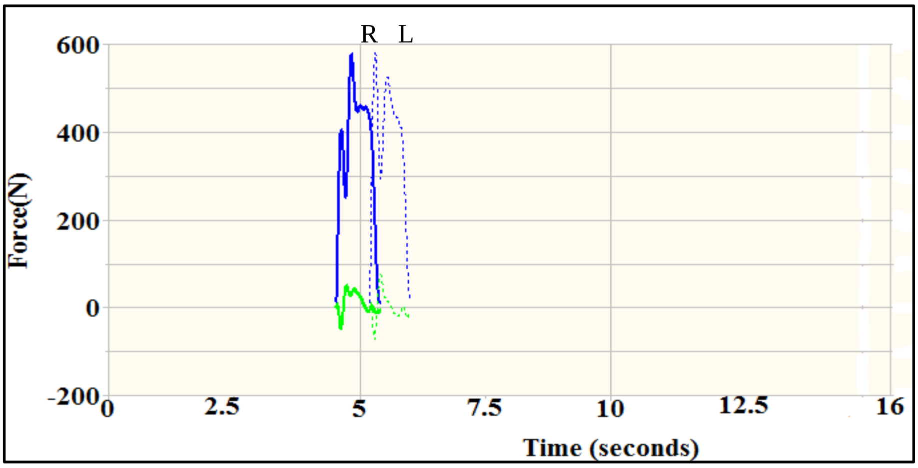


**Figure 5 Ground reaction force curve of a pathological gait**

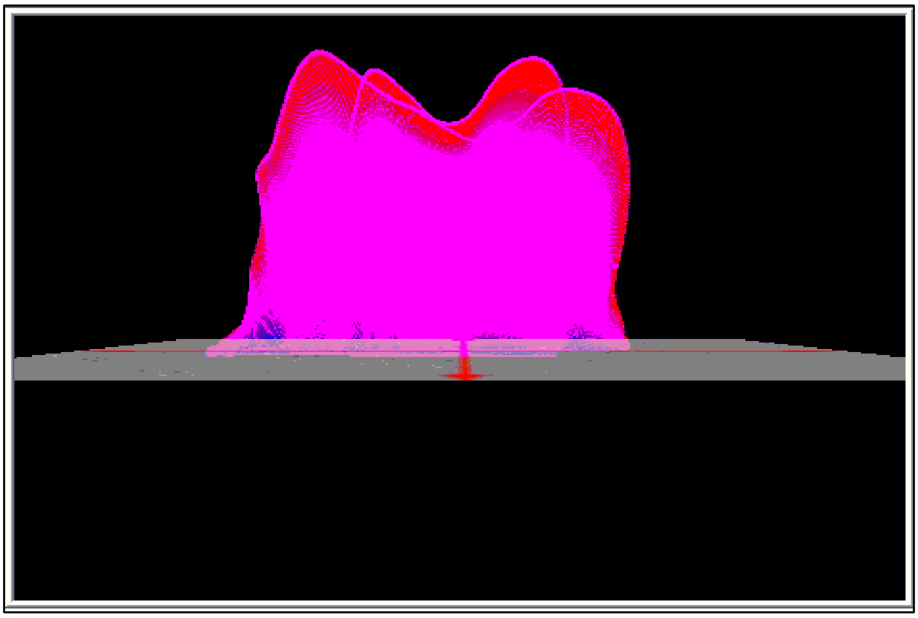

**Figure 6 Butterfly diagram of a ground reaction force curve.**

The subject has decelerated the swinging leg with less force magnitude (moment of ankle dorsiflexion is very small). Very low mid-stance force reveals unstability (risk of falling down, slipping etc) of the gait because postural regulation is coherently maintained by force applied by the individual as well as the reaction force exerted by the ground. Maximum force is exerted during propulsive phase when ankle begins to slowly plantar flexed; the force curve also starts to reduce due to the transition of body weight from one foot to the other. However, the gradual declining force curve is interrupted by a plateau region after the propulsive phase when it was supposed to attenuate completely. The ankle plantar flexors gets stretched (extends) and takes time in dragging the ground for acceleration. And this designates the difficulty in initiating the swing of the leg for acquiring required momentum of the body. The alternate foot following the planted foot signifies a better characteristic curve compared to the planted foot. In this case the gait pattern characterizes better functioning of the foot during impact loading and mid-stance phase. Here also a neck region arises during propulsive loading, it is noted that the subject is facing problems in lifting the leg due to weak Quadriceps muscle which is restricting the flexion of the knee joint.

Butterfly diagram shown in Figure 6 gives the corresponding force magnitude at every instant of time. It is known as butterfly diagram because of the butterfly shaped of set vectors. The butterfly diagram is also known as Pedotti diagram. From Pedotti diagram, vertical and horizontal forces are analysed for each moment in time. It is basically the discretized ground reaction force pattern use to quantify the force magnitude at each point from heel strike to toe-off phase.

### 4.1.2 Typical Centre of pressure curves of a normal and pathological gait

Centre of pressure curve is assessed for studying the body balance conditions while maintaining postural stability. A body is said to be in balance or mechanical equilibrium when the sum of all the forces and torques acting on the body both externally and internally is neutral or zero. The external forces are generally the gravitational force acting on the body and the ground reaction force acting on the plantar surface of the foot and internal forces are the heartbeats and muscular activations (physiological disturbances). During the time of erect posture, the gravitational and the ground reaction force does not show much variation, it remains almost constant but the internal forces cause by physiological responses are never constant, they continuously cause perturbations and when no external force acts on to it, body sway may occur depending upon the intensity of perturbations. It determines the variation along anterior-posterior and medial lateral direction. The typical curve of Centre of pressure is known as Stabilogram [17]. From Stabilogram the variation is observed only in a single plane whereas Statokinesigram defines the variation of body in both planes (anterio-posterior and medial-lateral). Centre of pressure is calculated according to Eqn.9. The centre of pressure curve (Statokinesigram) shown in Figure7 (A) represents the line of action upon which the gravitational force is acting. Figure7 (A) shows higher variation along medial-lateral direction during quiet standing with open eyes compared to anterior posterior direction which varies from the heel pad to the metatarsal heads. The Statokinesigram gives an indication that the line of action due to gravity acts laterally around the heel pad and moves diagonally towards the first and second metatarsal heads (big toe). Hence it is inferred that the subject's erect posture depends upon the base of support and is stable with line of action acting across the foot. In this Figure7 (A), it is observed that sudden perturbations occur in the forefoot section with large cluster covering a small area under curve revealing little/no sway. And shorter length of the cluster in the hind foot with larger area corresponds to fine control of the body balance.

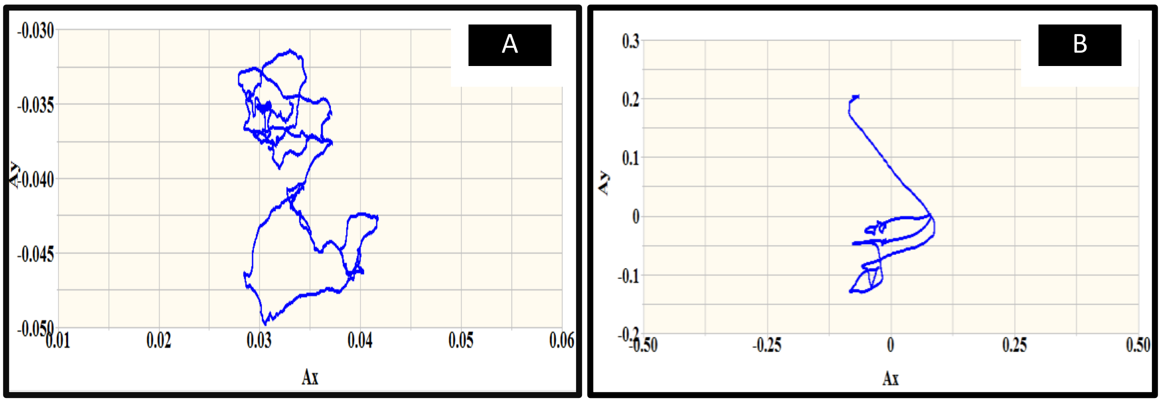


**Figure 7 Typical centre of pressure curve of A) normal and B) pathological gait**

From Figure 7(B) the centre of pressure graph is found to exhibit very high variation along medial-lateral direction rather than in anterior-posterior direction. It implies that the line of action is not acting along the entire foot sole. According to the graph the subject's whole load is concentrated around the heel and mid foot portion while standing. The subject's centre of pressure curve shows stability with perturbations and small cluster in the posterior part of the foot. This implies that the subject tends to fall forward due to the line of action acting along a single and shorter segment. And a longer duration in the same posture will affect the balance due to physiological disturbances caused by muscle activations and heartbeats that are continuously perturbing. The ground does not exert force once the plantar surface of the foot wholly interacts with the ground. In effect to the external force the internal force varying continuously disturbs the body balance condition leading to sway in both anterior-posterior and medial-lateral direction.

## 4.1.3 Typical Coefficient of friction graphs of a normal and pathological gait

The coefficient of friction graph for a normal healthy gait is shown in Figure 8 (A) during walking at self-selected speed. The coefficient of friction graph represents the pronation and supination of the foot during walking. From the Figure 8 (A) it is observed that while applying braking force, the foot (heel-pad) exhibits higher frictional force compared to that during toe-lifting for forward progression. Thus in case of normal gait it is observed that load acceptance phase by the foot faces higher frictional force and moves smoothly from the heel strike phase to the toe-off phase. From the graph it is also noted that the frictional force is higher than the normal force.

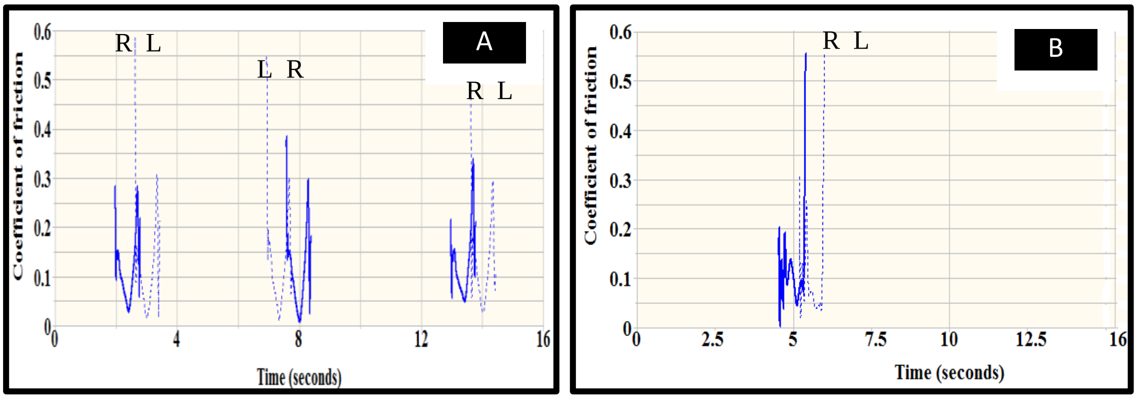


**Figure 8 Typical centre of pressure curve of A) normal and B) pathological gait**

In case of a pathologic gait shown in Figure 8 (B), the coefficient of friction is found to be higher during propulsive loading. This illustrates the lateral side of the planted foot is pronated highly which is limiting the subject’s foot to overcome the frictional force cause by the ground. As a result the subject takes longer duration to lift the foot for accelerating forward. During walking the heel pad slightly pushes back on the ground for decelerating the swinging motion of the leg and overcomes a certain amount of frictional force exerted by the ground. Similarly, with the ground reaction force and gravity acting on the plantar surface of the foot from heel pad to the metatarsal regions is found to be very fluctuating because every instant of the foot motion is affected by the weak and stiff muscles which do not contract and relax accordingly.

### 4.1.4 Torque

Torque acting at the joint represents the rotational effect of the bone joints for eccentric or concentric contractions. Torque acting clockwise is shown to be negative and the torque acting in counter-clockwise direction is considered to be positive. The torque of a normal subject is found to be higher at the time of heel strike when a braking force is applied to decelerate the motion of the swinging leg. Torque is calculated according to Eqn. 8. In all the gait cycles shown in the Figure9 (A), the torque at the time of heel strike is highest. The torque basically represents the rotation of the bone joint in response to the ground reaction force.

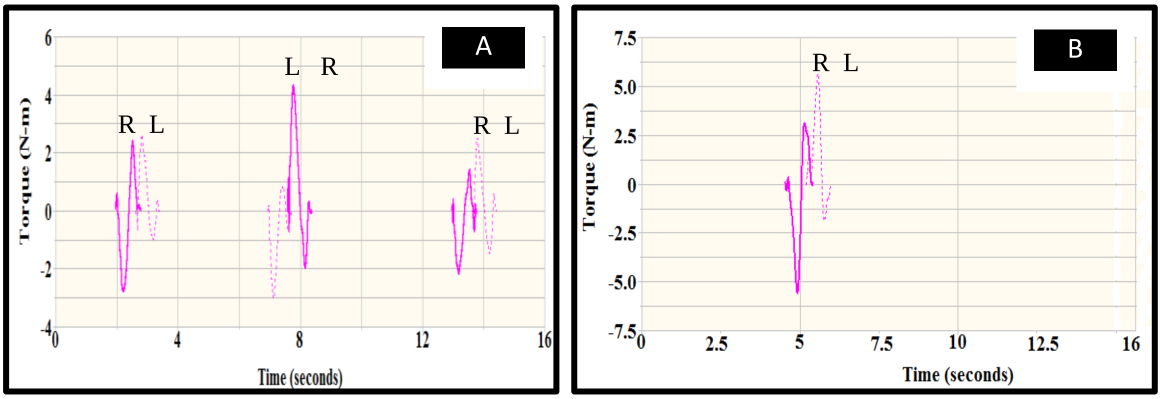


**Figure 9 Typical torque curve of A) normal and B) pathological gait**

The rotational nature of the bone joint in case of a pathological gait shown in Figure 9 reveals proper clockwise and counter-clockwise movement during deceleration and acceleration of the planted foot. However, the torque magnitude of a differently abled subject is two times higher than the normal healthy subject and this high rotational affect cause by ground reaction force during heel strike means that the foot is inverted. The foot inversion pressurizes the lateral borders of the foot and when it reaches the metatarsal regions, the inversion cause during heel strike laterally pronates the foot and results in longer pronation. Pronation in turn again depends on the musculo-skeletal system. Since, the muscles are stiff and weak, once the muscles get contracted, the muscles take time to relax and all these factors combination causes longer duration of pronation and hence increases the period to overcome the frictional force cause by the ground.

### 4.1.5 Power Dynamics

Power graph in gait analysis refers to the power executed by the joints in response to the movement patterns. The expression used for computing power is

$$P = F.V \qquad \ldots\ldots. \ (10)$$

where P represents power, F represents the respective force and V is the corresponding velocity. From Figure 10 (A) it is observed that output power is negative during decelerating the foot and with the change in motion of the leg the power output becomes positive after the midstance phase of the leg motion. In Figure 10 (A) a negative power is observed during heel strike and positive power after Midstance phase to propulsive phase. The negative power corresponds to the power absorption by the joints and positive power corresponds to the utilization of power for executing the motion. The power curve of a pathological gait shown in Figure 10 (B) shows no power absorption during heel strike, it only shows the utilization of power for executing the desired motion.

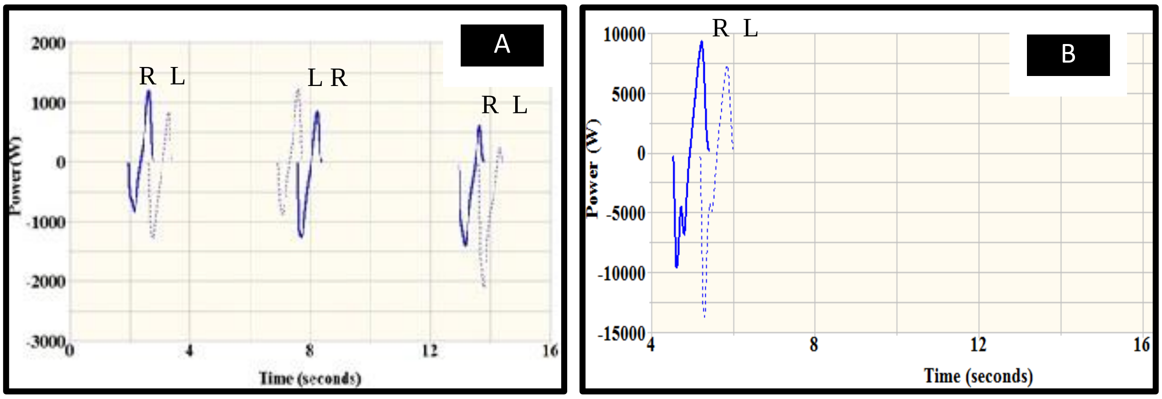


**Figure 10 Typical power curve of A) normal and B) pathological gait**

The power utilization is 100 times more than a normal subject shown in Figure 10 (A). This quantifies that the subject has problem in the front part of the foot, the subject takes large effort to lift the foot due to weak muscles and utilizes very high energy to full fill the desired motion. Due to spasticity the subject's joints are also flexed to certain degree and this restricts the range of motion. Due to flexed or bowed joints the power is not executed proportionally from one joint to the other and it directly reaches the forefoot of the ground creating the need for excess expenditure of power to overcome the lateral pronation cause by the muscle weakness.

### 4.1.6 Impulse force

The impulse force represents the stress applied during the motor act. Impulse is expressed as the integral of force with respect to time.

$$Impulse \ = \int_{T_1}^{T_2} J \qquad \ldots\ldots. \ (11)$$

Here J refers to the impulsive force and $T_1$ represents the time interval between initial contact and midstance phase and $T_2$ represents time interval between propulsive loading and toe-off phase. Impulse is very important in determining the functioning of hind foot and the forefoot. From the Figure 11(A) it is observed that with heel strike the impulse starts growing gradually because of the corresponding force and time. During this phase load transition takes place from double support stage to single support stage and the complete load of the body is applied on the single foot. Immediately after heel strike, the impulse starts increasing when the foot pronates to overcome the frictional force for lifting the foot.

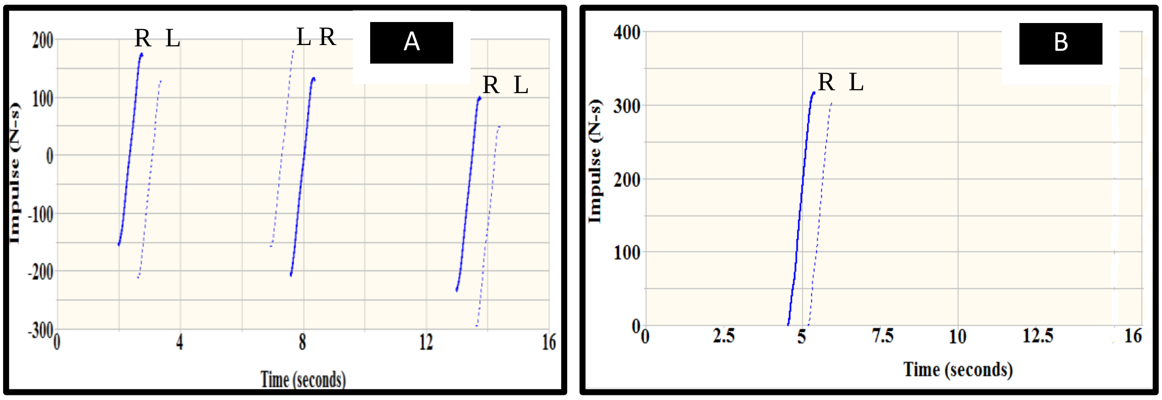


**Figure 11 Typical impulse curve of A) normal and B) pathological gait**

Similar to power curve, it is observed that the pathological gait shows high higher magnitude of torque in impulse curve shown in Figure11 (B). This reveals that the subject did not perform the three phases (heel strike, midstance and toe-off) of the gait during walking. Instead of heel strike, the subject had just slap on the ground and lifted the foot off the ground. From the typical performance parameters discussed in Figure 4 to Figure11 the difference between a normal and pathological gait is clearly specified. In all the performance parameters the pathological condition is observed carefully supported by quantitative and qualitative observations.

## 4.2 Dynamics of walking

### 4.2.1 Different phases of gait cycle

Walking is the most commonly performed gait, generally defined as the rhythmical pattern. It is very suitable and least expensive of all patterns of gait. The dynamics of walking is best understood with the help of the phases of gait cycle. The phases of gait cycle for 57 subjects including male, female and differently abled subjects during normal (self selected speed) walking are listed in Table 1. Out of 57 subjects 30 subjects are normal healthy male subjects, 24 are normal healthy female subjects and 3 subjects are differently abled (subject 1 to subject 54 are normal healthy subjects and subject 54 to subject 57 are differently abled). The type of disorder found in 3 differently abled subjects are described in section 3.1 under experimental methods.

**Table 1 Typical power curve of A) normal and B) pathological gait**

| Sl.no | Age | Height | Weight | Gender | Gait parameters during normal walking | | | | |
|---|---|---|---|---|---|---|---|---|---|
| | | | | | Duration of gait cycle (s) | Right stance phase (s) | Right swing phase (s) | Left swing phase(s) | Left stance phase(s) |
| **Subject 1** | 24 | 167.64 | 75 | Male | 1.155 | 0.762 | 0.414 | 0.513 | 0.743 |
| **Subject 2** | 22 | 167.64 | 50 | Female | 1.35 | 0.854 | 0.496 | 0.563 | 0.82 |
| **Subject 3** | 27 | 160 | 55 | Male | 1.294 | 0.816 | 0.472 | 0.536 | 0.763 |
| **Subject 4** | 26 | 183 | 70 | Male | 1.648 | 1.118 | 0.629 | 0.629 | 1.091 |
| **Subject** | 25 | 164.59 | 52 | Female | 1.166 | 0.722 | 0.444 | 0.502 | 0.742 |

| | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|
| **5** | | | | | | | | | |
| **Subject 6** | 25 | 162 | 55 | Female | 1.122 | 0.737 | 0.378 | 0.745 | 0.461 |
| **Subject 7** | 27 | 166 | 60 | Male | 1.395 | 0.873 | 0.521 | 0.566 | 0.919 |
| **Subject 8** | 28 | 159 | 60 | Male | 1.48 | 0.956 | 0.519 | 0.622 | 0.932 |
| **Subject 9** | 28 | 163 | 65 | Male | 1.25 | 0.803 | 0.447 | 0.479 | 0.802 |
| **Subject 10** | 24 | 161 | 51 | Female | 1.208 | 0.791 | 0.417 | 0.555 | 0.709 |
| **Subject 11** | 26 | 150 | 55 | Female | 1.398 | 0.924 | 0.465 | 0.649 | 0.844 |
| **Subject 12** | 27 | 165 | 67 | Male | 1.169 | 0.738 | 0.435 | 0.521 | 0.77 |
| **Subject 13** | 25 | 155 | 56 | Male | 1.072 | 0.684 | 0.391 | 0.468 | 0.558 |
| **Subject 14** | 28 | 160 | 61 | Male | 1.153 | 0.755 | 0.393 | 0.537 | 0.69 |
| **Subject 15** | 26 | 162 | 65 | Male | 1.317 | 0.87 | 0.447 | 0.848 | 0.604 |
| **Subject 16** | 25 | 185.92 | 68 | Male | 1.249 | 0.815 | 0.443 | 0.575 | 0.782 |
| **Subject 17** | 25 | 158.49 | 59 | Female | 1.138 | 0.741 | 0.397 | 0.513 | 0.732 |

| | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|
| **Subject 18** | 28 | 170.69 | 66 | Male | 1.22 | 0.8 | 0.42 | 0.768 | 0.525 |
| **Subject 19** | 27 | 164.59 | 69 | Male | 1.311 | 0.901 | 0.41 | 0.565 | 0.821 |
| **Subject 20** | 28 | 165 | 70 | Male | 1.245 | 0.806 | 0.439 | 0.538 | 0.806 |
| **Subject 21** | 27 | 166 | 67 | Male | 1.295 | 0.877 | 0.418 | 0.513 | 0.813 |
| **Subject 22** | 25 | 168 | 72 | Male | 1.171 | 0.787 | 0.384 | 0.547 | 0.741 |
| **Subject 23** | 29 | 173 | 75 | Male | 1.311 | 0.843 | 0.468 | 0.576 | 1.424 |
| **Subject 24** | 27 | 166 | 65 | Male | 1.038 | 0.657 | 0.381 | 0.451 | 0.642 |
| **Subject 25** | 28 | 176.78 | 67 | Female | 1.433 | 0.962 | 0.471 | 0.604 | 0.895 |
| **Subject 26** | 23 | 170.68 | 66 | Female | 1.033 | 0.674 | 0.359 | 0.456 | 0.657 |
| **Subject 27** | 26 | 186 | 78 | Male | 1.38 | 0.923 | 0.457 | 0.586 | 0.888 |
| **Subject 28** | 28 | 176 | 68 | Male | 1.073 | 0.691 | 0.382 | 0.478 | 0.647 |
| **Subject 29** | 29 | 160 | 66 | Male | 1.285 | 0.856 | 0.429 | 0.881 | 0.561 |
| **Subject** | 26 | 155 | 52 | Female | 1.453 | 0.951 | 0.502 | 0.613 | 0.897 |

| | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|
| **30** | | | | | | | | | |
| **Subject 31** | 27 | 161 | 59 | Male | 1.183 | 0.763 | 0.42 | 0.526 | 0.749 |
| **Subject 32** | 29 | 160 | 64 | Female | 1.344 | 0.892 | 0.452 | 0.579 | 0.869 |
| **Subject 33** | 28 | 166 | 72 | Male | 1.526 | 0.892 | 0.634 | 0.549 | 0.875 |
| **Subject 34** | 26 | 182 | 74 | Male | 1.178 | 0.753 | 0.425 | 0.506 | 0.768 |
| **Subject 35** | 27 | 178 | 79 | Male | 1.121 | 0.703 | 0.427 | 0.522 | 0.726 |
| **Subject 36** | 26 | 174 | 48 | Female | 1.178 | 0.753 | 0.425 | 0.713 | 0.528 |
| **Subject 37** | 25 | 178 | 69 | Male | 1.262 | 0.833 | 0.429 | 0.579 | 0.835 |
| **Subject 38** | 28 | 175 | 67 | Male | 1.26 | 1.1 | 0.16 | 0.488 | 1.028 |
| **Subject 39** | 28 | 171 | 57 | Female | 1.209 | 0.762 | 0.447 | 0.488 | 0.792 |
| **Subject 40** | 31 | 175 | 65 | Male | 1.054 | 0.734 | 0.32 | 0.692 | 0.471 |
| **Subject 41** | 24 | 175 | 64 | Male | 1.152 | 0.736 | 0.416 | 0.507 | 0.772 |
| **Subject 42** | 25 | 172 | 49 | Female | 0.977 | 0.61 | 0.367 | 0.43 | 0.623 |

| | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|
| **Subject 43** | 26 | 171 | 63 | Female | 1.128 | 0.724 | 0.404 | 0.483 | 0.712 |
| **Subject 44** | 25 | 170 | 52 | Male | 1.722 | 1.148 | 0.574 | 0.754 | 1.062 |
| **Subject 45** | 30 | 155 | 56 | Female | 2.214 | 0.785 | 1.429 | 0.568 | 0.765 |
| **Subject 46** | 27 | 157 | 58 | Male | 1.214 | 0.801 | 0.413 | 0.541 | 0.784 |
| **Subject 47** | 28 | 165 | 67 | Male | 1.754 | 0.997 | 0.757 | 0.647 | 0.963 |
| **Subject 48** | 26 | 169 | 58 | Female | 0.988 | 0.657 | 0.386 | 0.475 | 0.664 |
| **Subject 49** | 25 | 173 | 65 | Male | 1.318 | 0.868 | 0.45 | 0.587 | 0.818 |
| **Subject 50** | 26 | 177 | 60 | Male | 1.608 | 0.992 | 0.614 | 0.73 | 0.981 |
| **Subject 51** | 28 | 174 | 64 | Male | 1.06 | 0.662 | 0.398 | 0.513 | 0.667 |
| **Subject 52** | 27 | 178 | 75 | Male | 1.133 | 0.709 | 0.495 | 0.447 | 0.624 |
| **Subject 53** | 27 | 176 | 72 | Male | 1.151 | 0.75 | 0.401 | 0.552 | 0.726 |
| **Subject 54** | 25 | 180 | 78 | Male | 1.164 | 0.738 | 0.434 | 0.549 | 0.766 |
| ***Subject** | 24 | 165 | 52 | Male | 1.34 | 0.852 | 0.384 | 0.576 | 0.804 |

| | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|
| **55** | | | | | | | | | |
| ***Subject 56** | 24 | 169 | 58 | Male | 1.733 | 1.067 | 0.666 | 0.845 | 1.057 |
| ***Subject 57** | 27 | 172 | 57 | Male | 1.955 | 1.297 | 0.658 | 0.761 | 1.247 |

***Subject 55 – Spastic Quadriplegia, *Subject 56 – Orthopedically handicapped 50%, *Subject 57 – Ataxic gait.**

From the Table 1 it is observed that gait cycle duration ranges from $1.01 \pm 1.2$. Lower duration of gait cycle infers high speed gait pattern and longer duration infers slow gait pattern. Among 57 subjects (normal male and female subject and differently abled subject) none of the subjects exhibit any severe pain or injury during their gait activities.

By calculating the average values of different phases of the gait cycle it is noted that right stance phase or first planted foot holds maximum duration in a single gait cycle. This implies that the first planted foot plays major role in maintaining the motion of the foot. Right stance phase is followed by left swing phase which again holds almost equivalent duration like the initial planted foot.

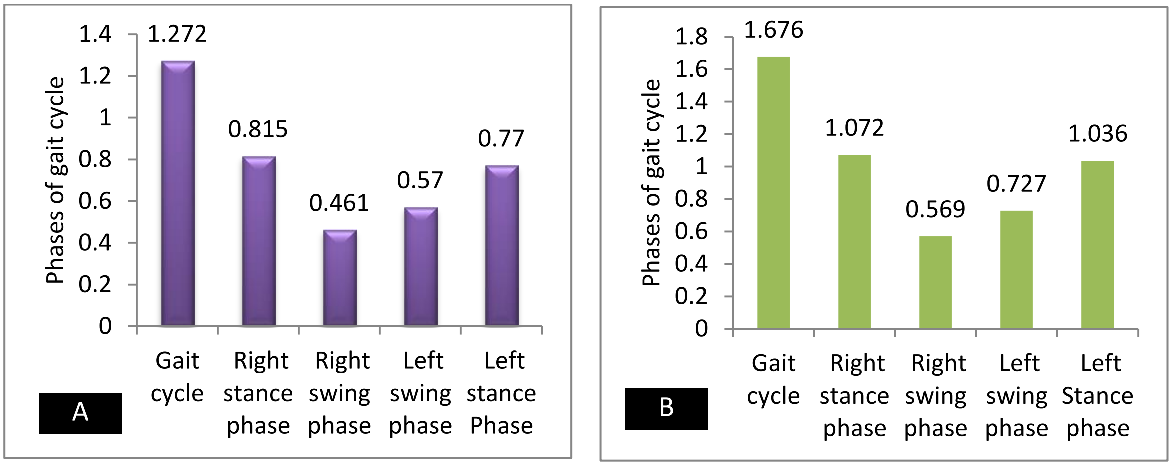


**Figure 12 Different phases of gait cycle for A) normal and B) pathological gait**

Right swing phase occurs for least duration in the gait cycle because of the traditional walking method following the general pattern of walking. When the mean variations are compared for normal shown in Figure 12 (A) and differently abled subjects shown in

Figure 12 (B) it is observed that pathological gait phases exhibited by differently abled subjects require higher time duration to perform the gait activities.

## 4.3 Analysing the gait patterns at different rates

### 4.3.1 Self-selected speed gait pattern

The gait patterns exhibited by a normal healthy subject are found to be same with that in Figure 4 and Figure 6. For analysing the gait patterns at different rates, self-selected speed is first chosen in order to observe the natural gait parameters which is not deteriorated by any kind of variations. Once the subjects are instructed their gait pattern will vary and it would not be possible to compare different phases of gait cycle at different rates. Moreover, the self-selected speed gait is already referred in Figure 4 and Figure 6, so it is not represent here again. Hence, only the index of forefoot function is calculated to observe that the front foot plays the superior role in gait motion. In order to observe the effect of load in the foot quantitatively the index of forefoot function is computed as given by M. Grundy [20]

$$Index\ of\ forefoot\ function = \frac{Forefoot\ impulse}{Hindfoot\ impulse}$$

$$Index\ of\ forefoot\ function\ (for\ each\ step) = \frac{\int_{t_1}^{t_2} F_f\, dt}{\int_{t_1}^{t_2} F_h dt} \quad \text{........ (12)}$$

**Table 2 Forefoot function index of 57 subjects walking at their self-selected speed**

| Sl.no | Forefoot function index during walking at self-selected speed (N-m) |
|---|---|
| **Subject 1** | 0.703 |
| **Subject 2** | 0.945 |
| **Subject 3** | 0.842 |
| **Subject 4** | 0.89 |
| **Subject 5** | 1.083 |
| **Subject 6** | 1.246 |
| **Subject 7** | 0.902 |

| | |
|---|---|
| **Subject 8** | 1.072 |
| **Subject 9** | 0.731 |
| **Subject 10** | 0.944 |
| **Subject 11** | 0.981 |
| **Subject 12** | 1.02 |
| **Subject 13** | 1.042 |
| **Subject 14** | 0.758 |
| **Subject 15** | 1.074 |
| **Subject 16** | 0.996 |
| **Subject 17** | 0.943 |
| **Subject 18** | 1.064 |
| **Subject 19** | 0.943 |
| **Subject 20** | 1.47 |
| **Subject 21** | 0.79 |
| **Subject 22** | 1.065 |
| **Subject 23** | 0.895 |
| **Subject 24** | 0.875 |
| **Subject 25** | 0.723 |
| **Subject 26** | 1.031 |
| **Subject 27** | 0.818 |
| **Subject 28** | 0.858 |
| **Subject 29** | 1.083 |
| **Subject 30** | 0.919 |
| **Subject 31** | 1.0277 |
| **Subject 32** | 0.921 |
| **Subject 33** | 0.974 |

| | |
|---|---|
| **Subject 34** | 1.01 |
| **Subject 35** | 0.964 |
| **Subject 36** | 0.807 |
| **Subject 37** | 1.145 |
| **Subject 38** | 1.3 |
| **Subject 39** | 0.761 |
| **Subject 40** | 0.931 |
| **Subject 41** | 1.083 |
| **Subject 42** | 0.851 |
| **Subject 43** | 0.861 |
| **Subject 44** | 1.031 |
| **Subject 45** | 1.056 |
| **Subject 46** | 1.017 |
| **Subject 47** | 0.852 |
| **Subject 48** | 0.87 |
| **Subject 49** | 0.584 |
| **Subject 50** | 1.047 |
| **Subject 51** | 1.275 |
| **Subject 52** | 0.855 |
| **Subject 53** | 0.79 |
| **Subject 54** | 1.123 |
| ***Subject 55** | 5.694 |
| ***Subject 56** | 1.214 |
| ***Subject 57** | 0.77 |

***Subject 55 – Spastic Quadriplegia, *Subject 56 – Orthopedically handicapped 50%, *Subject 57 – Ataxic gait.**

### 4.3.2 Slow gait pattern

In this gait pattern a large variation is observed compared to the normal gait pattern. The symmetric humps in the curves are missing in the ground reaction force curve implying reduced stress during walking gait of a normal subject. Accordingly gait must be performed at ease with minimum expenses.

As the subjects do not perform it usually (slower than their normal self-selected speed) the stress reduction in first step length is not observed for the second step length shown in Figure 13 (A).

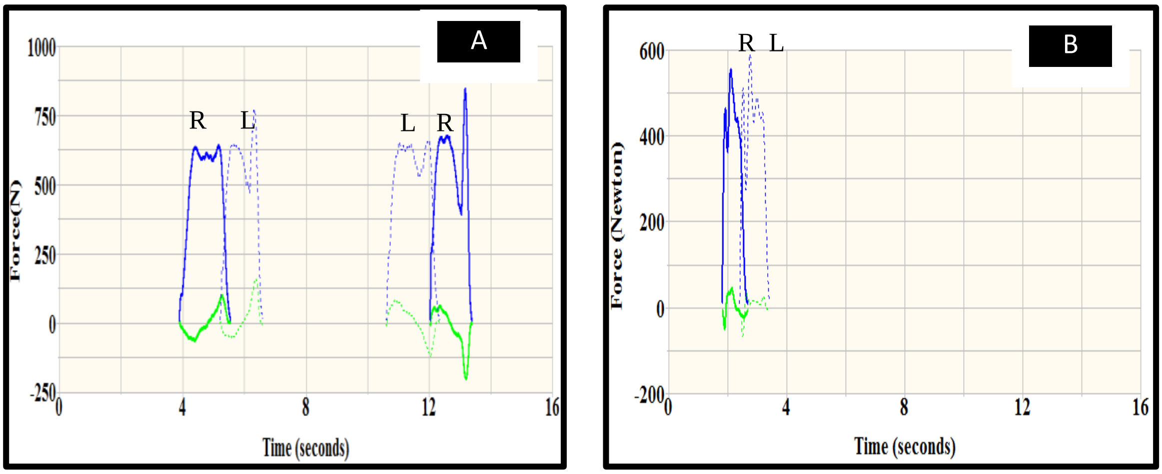


**Figure 13 Slower gait patterns of A) normal and B) differently abled subject**

In case of slow gait, the normal subjects initially apply less stress; but during transition of single support phase to double support phase [12-14] the stress increases highly. When the slow gait is observed for differently abled subject shown in figure 13 (B), the stress reduction is not observed unlike in normal subject Figure 13 (A). Instead sharp peaks were observed and this is due to the uncoordinated muscle contraction and relaxation. This causes higher impulse at mid-stance phase and lower impulse at propulsive phase, which means that the tendon and muscles connecting the leg and foot are contracted (stretched) but does not relax with immediate contraction.

**Table 3 Forefoot function index of 57 subjects walking at slower speed than their self-selected speed**

| Sl. no | Forefoot function index during slow gait than their self-selected speed (N-m) |
|---|---|
| **Subject 1** | 0.77 |
| **Subject 2** | 0.876 |
| **Subject 3** | 0.682 |
| **Subject 4** | 0.797 |
| **Subject 5** | 0.832 |
| **Subject 6** | 0.914 |
| **Subject 7** | 0.926 |
| **Subject 8** | 0.772 |
| **Subject 9** | 0.872 |
| **Subject 10** | 0.79 |
| **Subject 11** | 0.666 |
| **Subject 12** | 0.969 |
| **Subject 13** | 0.805 |
| **Subject 14** | 0.833 |
| **Subject 15** | 1.06 |
| **Subject 16** | 1.217 |
| **Subject 17** | 1.067 |
| **Subject 18** | 0.68 |

| | |
|---|---|
| **Subject 19** | 0.679 |
| **Subject 20** | 0.792 |
| **Subject 21** | 0.976 |
| **Subject 22** | 0.788 |
| **Subject 23** | 1.1 |
| **Subject 24** | 0.875 |
| **Subject 25** | 0.78 |
| **Subject 26** | 0.79 |
| **Subject 27** | 0.85 |
| **Subject 28** | 0.95 |
| **Subject 29** | 1.32 |
| **Subject 30** | 0.70 |
| **Subject 31** | 0.899 |
| **Subject 32** | 0.953 |
| **Subject 33** | 0.754 |
| **Subject 34** | 0.972 |
| **Subject 35** | 0.769 |
| **Subject 36** | 0.82 |
| **Subject 37** | 1.2 |
| **Subject 38** | 0.938 |
| **Subject 39** | 0.685 |

| | |
|---|---|
| **Subject 40** | 0.75 |
| **Subject 41** | 0.65 |
| **Subject 42** | 0.945 |
| **Subject 43** | 0.92 |
| **Subject 44** | 0.935 |
| **Subject 45** | 0.954 |
| **Subject 46** | 1.03 |
| **Subject 47** | 0.8 |
| **Subject 48** | 0.74 |
| **Subject 49** | 0.87 |
| **Subject 50** | 1.109 |
| **Subject 51** | 0.95 |
| **Subject 52** | 1.09 |
| **Subject 53** | 0.89 |
| **Subject 54** | 0.99 |
| ***Subject 55** | 1 |
| ***Subject 56** | 0.876 |
| ***Subject 57** | 0.98 |

***Subject 55 – Spastic Quadriplegia, *Subject 56 – Orthopedically handicapped 50%, *Subject 57 – Ataxic gait.**

### 4.3.3 Fast gait pattern

When the fast gait pattern is observed (faster than their normal gait pattern) high impulses resulting in large stress is noted. Two- hump symmetric curve is again observed shown in Figure 14 (A) and that too is comparatively sharper than their normal speed gait. However, this gait pattern is not desirable because, it utilizes high energy to execute making it more expensive.

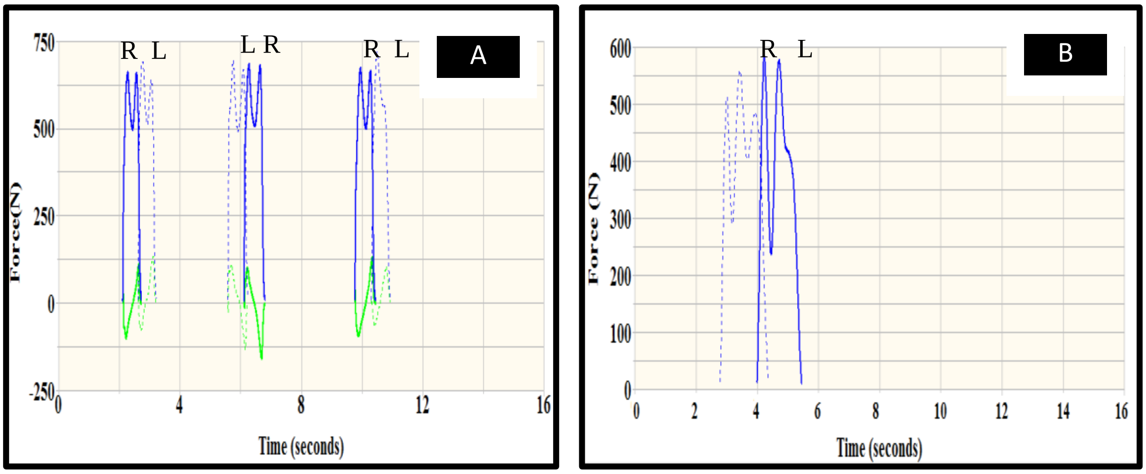


**Figure 14 Faster gait patterns of A) normal and B) pathological gait**

When the phases of gait cycle are computed it is observed that impact loading and mid-stance phase are comparable to a normal healthy subject but propulsive loading shows large time utilization and less percentage of gait cycle performed. In case of normal healthy subject the impulse grows higher with gait rate variation but in case of differently abled subject, the variation of the gait rate shows less percentage of gait phases performed.

**Table 4 Forefoot function index of 57 subjects walking at faster speed**

| Sl. no | Forefoot function index during fast gait their self-selected speed (N-m) |
|---|---|
| **Subject 1** | 0.812 |
| **Subject 2** | 0.912 |
| **Subject 3** | 0.934 |

| | |
|---|---|
| **Subject 4** | 1.04 |
| **Subject 5** | 0.967 |
| **Subject 6** | 0.893 |
| **Subject 7** | 0.933 |
| **Subject 8** | 0.747 |
| **Subject 9** | 0.74 |
| **Subject 10** | 0.75 |
| **Subject 11** | 0.725 |
| **Subject 12** | 0.822 |
| **Subject 13** | 0.834 |
| **Subject 14** | 0.728 |
| **Subject 15** | 1.05 |
| **Subject 16** | 1.026 |
| **Subject 17** | 0.98 |
| **Subject 18** | 0.866 |
| **Subject 19** | 0.736 |
| **Subject 20** | 0.675 |
| **Subject 21** | 1.27 |
| **Subject 22** | 0.863 |
| **Subject 23** | 0.765 |
| **Subject 24** | 0.657 |

| | |
|---|---|
| **Subject 25** | 0.839 |
| **Subject 26** | 0.864 |
| **Subject 27** | 0.85 |
| **Subject 28** | 0.76 |
| **Subject 29** | 0.93 |
| **Subject 30** | 1.08 |
| **Subject 31** | 0.958 |
| **Subject 32** | 0.951 |
| **Subject 33** | 1.09 |
| **Subject 34** | 1 |
| **Subject 35** | 1.037 |
| **Subject 36** | 0.89 |
| **Subject 37** | 1.13 |
| **Subject 38** | 1.95 |
| **Subject 39** | 0.617 |
| **Subject 40** | 0.84 |
| **Subject 41** | 1.14 |
| **Subject 42** | 0.885 |
| **Subject 43** | 0.606 |
| **Subject 44** | 0.95 |
| **Subject 45** | 1.114 |

| | |
|---|---|
| **Subject 46** | 1.07 |
| **Subject 47** | 0.8 |
| **Subject 48** | 1.31 |
| **Subject 49** | 0.95 |
| **Subject 50** | 1.02 |
| **Subject 51** | 0.87 |
| **Subject 52** | 1.25 |
| **Subject 53** | 1.28 |
| **Subject 54** | 3.04 |
| ***Subject 55** | 2.31 |
| ***Subject 56** | 1.99 |
| ***Subject 57** | 2.1 |

***Subject 55 – Spastic Quadriplegia, *Subject 56 – Orthopedically handicapped 50%, *Subject 57 – Ataxic gait.**

By observing the gait rates for different subjects it is inferred that the normal gait pattern (self- selected gait pattern) the forefoot index function is highest. This means that the general walking pattern exhibited by these 57 subjects applies high impulse on the forefoot. After instruction the modified gait readings are observed and found that the forefoot function index is slower than that in normal pattern. Thus it is inferred from this analysis that in normal gait pattern shown in Figure 15 (A) high stress is applied which can increase the cost of walking by utilizing higher energy and power.

When this pattern is maintained for walking a longer distance, the subjects fatigue will get weakened and they will not be able to perform the work magnificently. However in case of pathological gait observed in Figure 15 (B) forefoot function index is highest in slower gait pattern followed by self-selected and fast gait. This implies that the differently abled subjects are facing difficulty in changing the gait pattern from self-selected speed to slow speed.

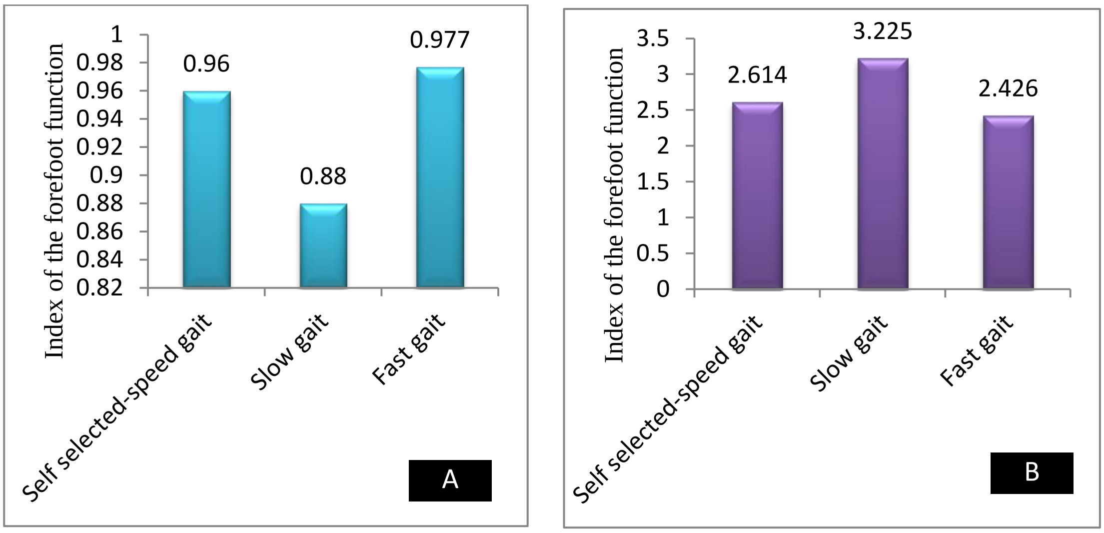


**Figure 15 Forefoot function index of A) normal and B) pathological gait**

Since they face problem in lifting their foot at their normal speed at slower speed the problem is enhanced holding more time duration to execute the desired gait. In case of fast gait, they invest more energy to achieve the required gait showing minimum forefoot function. So, from the analysis it is recommended that the subject normal subject's should try to perform a slower gait pattern than their normal pattern in order to perform the work for a longer duration without fatigue. And for differently abled subject according to the observation they must try to walk faster than their normal pattern as it acquires least forefoot function.

## 4.4 Analysis of ankle-joint forces during walking gait pattern

Walking is a rhythmic pattern showing twin-peak curve where the ankle joint plays the role of plantarflexion and dorsiflexion along with inversion and eversion. Irrespective of the entire body structure and functions, the lower extremity of human beings is considered to be most effective for locomotion. Hip-joint, knee-joint and ankle joint contribute to the goal of desired motional activities whereas proper regulation of these crucial joints is necessary for living a healthy life. Out of the three crucial joints, the ankle motion executes the desired motional activities employing the functions of the entire body [23]. The ankle-joint is a bone joint specifically termed as a hinge joint, located in the anterior part of the leg with 26 bones and 57 joints. The ankle-joint allows motion only in a single plane (up and down movement) [8] . The ankle-joint along with the foot form a complex of functional parts (talocrural joint, subtalar joint, transverse joint, tarsometatarsal joint, metatarsophalangeal joint and interphalangeal joint) enabling a person to perform a motor act like standing, walking, running and jumping.

**Table 5 Parameters of gait cycle**

| Self-selected speed gait | Parameters | Normal healthy subject | Corresponding % of gait cycle | Differently abled subject | % of gait cycle |
|---|---|---|---|---|---|
| | **Gait cycle 100% (s)** | 1.233 | 100 | 1.22 | 100 |
| | ***DF(N)** | 308.66 | 4.78 | 192.77 | 3 |
| | ***MSP(N)** | 642.18 | 19.54 | 400.74 | 7.21 |
| | ***PF(N)** | 669.01 | 48.74 | 577.83 | 24.26 |
| **Slower than self-selected speed gait** | **Gait cycle (s)100%** | 2.343 | 100 | 1.312 | 100 |
| | **DF(N)** | 88.62 | 1.7 | 169.07 | 2.13 |
| | **MSP(N)** | 632.1 | 21.51 | 463.08 | 8.53 |
| | **PF(N)** | 639.62 | 54.24 | 554.17 | 22.25 |
| **Faster than the self-selected speed gait** | **Gait cycle (s)100%** | 0.967 | 100 | 1.116 | 100 |
| | **DF(N)** | 100.43 | 54.8 | 268.35 | 3.58 |
| | **MSP(N)** | 651.14 | 67.32 | 508.75 | 10.03 |
| | **PF(N)** | 646.42 | 97.31 | 561.12 | 22.636 |

***DF – Dorsiflexion, *MSP – Mid stance phase, *PF – Plantarflexion**

The ankle-joint alone is not enough to perform motion in all directions, hence it is accompanied by a subtalar-joint (allows side-by-side motion), when combined together they act as a universal joint and participates in supination, pronation, dorsiflexion, plantarflexion, adduction and abduction. When the phases of gait cycle are computed it is observed that impact loading and mid-stance phase are comparable to a normal healthy subject but propulsive loading shows large time utilization and less percentage of gait cycle performed. In case of normal healthy subject the impulse grows higher with gait rate variation but in case of differently abled subject, the variation of the gait rate shows less percentage of gait phases performed.

From the Table 5, the mean for a single support phase of a normal healthy subject is computed as 0.483 seconds and double support phase holds 0.156 second for three gait cycles. The mean step length is calculated as 0.639 seconds for a gait cycle of 1.233 seconds. In case of differently abled subject, the single support phase holds for 0.408 seconds and double support phase holds for 0.212 seconds in a single gait cycle with step length of $(0.6 \pm 0.044)$ in a gait cycle of 1.22 seconds. Basically leg muscle and tendons are responsible for providing energy required to accelerate and decelerate the motion of the leg. While performing flexion and extension the leg muscles shortens and lengthens accordingly depending upon the gait activities. However the tendons acts like springs exhibiting elastic properties. And Achilles tendon which connects calf to the heel provides required strain energy for stretching the foot to perform gait. From Table 5 that a normal healthy subject walking at self-selected speed has performed 4.78% of gait during dorsiflexion and 19.54% of gait during midstance phase and 48.74% of gait during plantar flexion. When the same is observed for a differently abled subject it is found that the subject has performed 3% of gait during dorsiflexion and 7.21% of gait during midstance phase and 24.26% of gait during plantar flexion at self-selected speed. Thus, it implies that the gait of a differently able subject is almost half a time less than the normal able subject. Dorsiflexion is quite normal, but mid stance phase and plantar flexion shows very less amount of gait performed indicating the difficulty facing by the subject's foot in load distribution from hind foot to the anterior part of the foot. And this flexibility limitation is attributed to the function of Achilles tendon, plantar flexor muscles and dorsiflexor muscles. When the ankle-foot relations for slower and faster gait is observed, a normal subject shows less percentage of gait performed in more time during slow gait, optimum % of gait in self-selected speed gait and high percentage of gait performed in less time during fast walking. Whereas a differently able subject shows non-linear relation for slow, normal and fast gait pattern indicating the high time required for clearing the ground.

### 4.4.1 Analysing centre of mass acceleration, power and centre of pressure along anterior-posterior direction for postural balance at self-selected speed

The centre of mass acceleration is found to be increasing with heel strike because of the foot accelerating forward and becomes negative after midstance and propulsive phase. The average acceleration of a subject with weight 650 Newton has 0.795 $m/s^2$ during normal walking shown in Figure 16. The average power of the braking force is calculated and found to be (-1440.576Watt) shown in Figure10 (A). The braking force power is nothing but the ankle dorsiflexor power and propulsive force utilizes average power of (+911.82 Watt) which is actually ankle plantar flexor power occurring during push-off stage. Experimentally it is observed that with walking speed the power difference tends to increase [15].The power utilization depends upon the functions of distal muscles; this gets clearer when it is seen that the characteristic power graph of a differently abled subject walking at self-selected speed. Here it is observed that the net ankle dorsiflexor powers is (-876.54) and ankle plantar flexor power is (+ 823.8) shown in Figure10 (A). During ankle dorsiflexion much fluctuation is observed whereas ankle plantar flexors show smooth peaks. This corresponds again to the limited flexibility of the tendon joining the leg and foot i.e. Achilles tendon [14], due to which the subject is unable to apply the load immediately after the heel strike and also stiff and tone muscles are also equally acquainted. The centre of pressure is one of the most important determinants of gait [17]. Centre of pressure is used to determine the postural stability of gait activities performed by living beings [9].

In this experiment we analyse the centre of pressure along the anterio-posterior direction and observed that the line of action upon which the gravity is acting is deviating laterally for both the foot in case of a normal healthy subject. And for a differently abled subject also, the same principle is valid, where the line of action deviates laterally for the respective foot. Centre of pressure curves assess the postural control system for balancing the gait. From the lowest section of the graph in Figure17 it is observed that the line of action

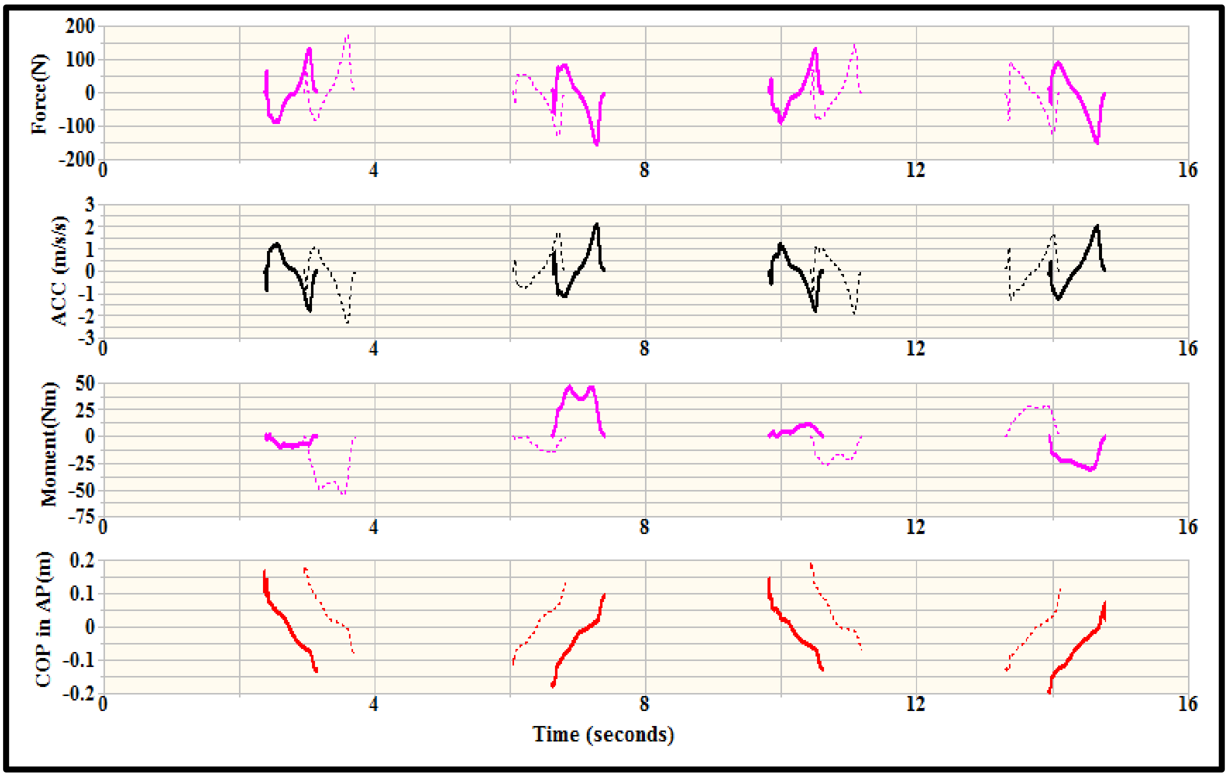


**Figure 16 Profiles of ground reaction force pattern, acceleration, moment and centre of pressure along y direction exhibited by normal healthy subject**

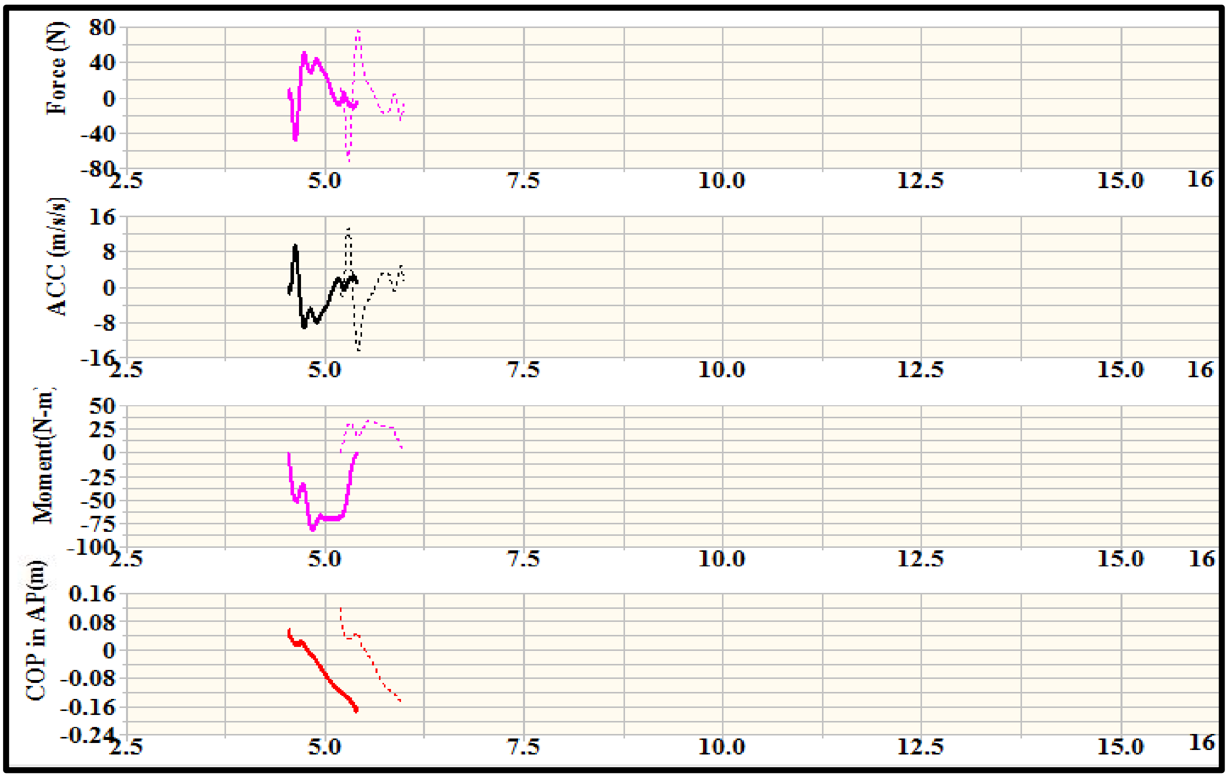


**Figure 17 Profiles of ground reaction force pattern, acceleration, moment and centre of pressure along y direction exhibited by a differently abled subject.**

initiates from the midline of the heel pad and then progresses towards the lateral direction of the foot i.e. along the metatarsophalanges. The line of action initially provides stability and with the motion of the leg it deviates from the central line at the time of execution of the body for diverse movements. Sometimes fall may occur due to misbalancing of the internal and external force. The external force is only the gravitational force and it exerts only when an internal force is applied. Internal force arises from the perturbations cause by muscles and physiological disturbances cause by respiration, heartbeats etc. Human body is never in equilibrium because of the internal force perturbations and when the body does not move, no external force acts to balance the system and as a result the body tend to fall. From figure 10 we observed that the net centre of mass acceleration (acceleration= $0.003 m/s^2$) is relatively high when compared to normal healthy subject (acceleration=$0.51 m/s^2$). The high centre of mass acceleration for differently abled subject is due to the muscle perturbations. Thus, the differently abled subject comprising of stiff and tone muscle if could not apply the load to the ground properly, the postural balance system will get deteriorated causing into fall of the subject. To avoid fall the subject must try to activate the leg and foot muscle. The last second graphs in Figure 16 and Figure17 correspond to the moment variation with time. It is observed that the rotational effect varies with the direction of moment applied showing negative phase during heel-strike or dorsiflexion and positive phase during plantar flexion. This trend is valid in both the case normal healthy subject and differently abled subject.

## 4.5 Analysis of Barefoot and shod-walking

From the investigation of barefoot and shod walking, it is inferred that when a subject walks in barefoot condition it exerts higher magnitude of force and the stress applied upon the heel pad and forefoot has significant effect.

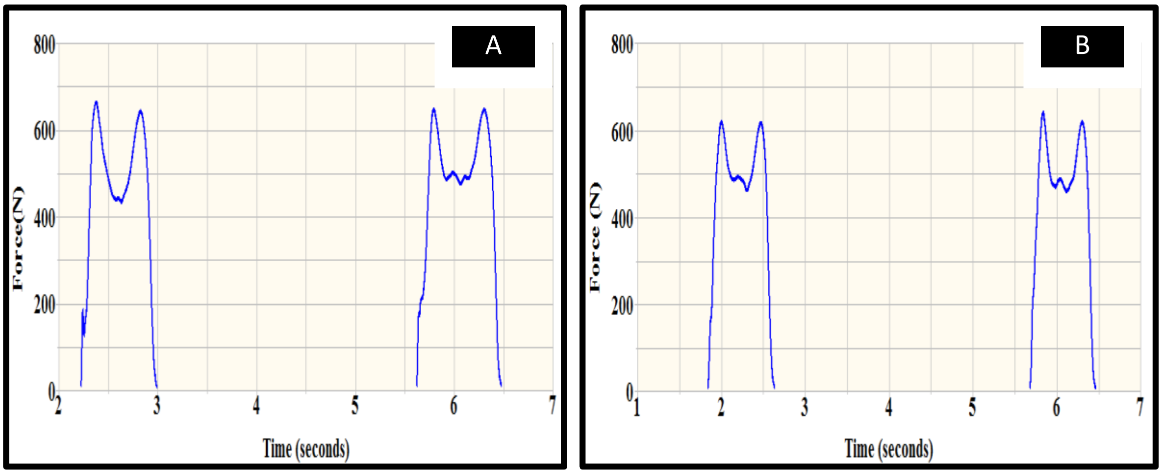


**Figure 18 A ground reaction force curve obtained during A) barefoot and B) shod walking of a normal subject.**

Though the pattern of ground reaction force (M-curve) is maintained shown in Figure 18, from the numerical values it is found that barefoot walking requires more fore utilization due to the direct contact of the plantar surface of the foot with the ground, where larger area of the foot comes in contact with the ground and the heel pad transfers the load by pronating laterally towards the forefoot. However when shod walking is observed as in Table 6, the numerical values show variation in force utilization. The force reduction observed is due to the shoe heel rigidity which absorbs the force and support in propagating the load towards the forefoot. In this case the function of the forefoot is also reduced to certain extent.

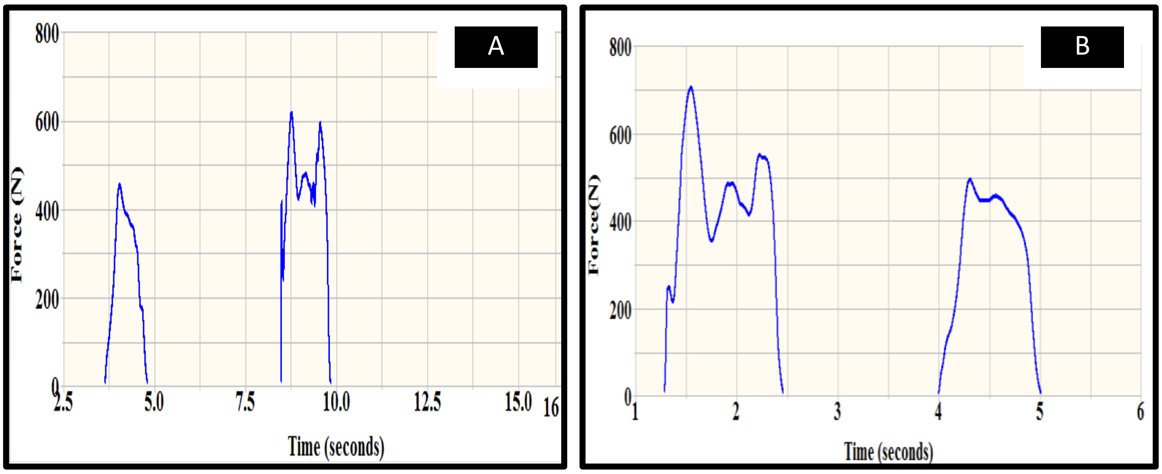


**Figure 19 A ground reaction force curve obtained during A) barefoot and B) shod walking of differently abled subject.**

**Table 6 Average force magnitude values for barefoot and shod walking**

| Type of gait | | Force magnitude (N) |
|---|---|---|
| 1) Normal gait | Barefoot | 176 |
| | Shod | 154 |
| 2)Pathological gait | Barefoot | 149 |
| | Shod | 230 |

The forefoot function is quantitatively observed by measuring the time duration of the forefoot contact with the ground.

## 4.6 Dynamics of Jumping

The dynamics of vertical jump is expressed by using flight height method [16]. With the help of law of conservation of mechanical energy during the flight phase, the relation between flight height and the velocity is obtained. While jumping vertically the resistance cause by air is neglected because of its minimal obstruction. With negligible air resistance, the vertical height is considered as a projectile in the flight phase. Considering the kinetic energy and gravitational potential energy during the time of take-off as $t_o$ and $t_{peak}$ when the vertical jump is at peak value the equation is given as

$$\frac{1}{2}mv_{t_o}^2 + mgh_{t_o} = \frac{1}{2}mv_{t_{peak}}^2 + mgh_{t_{peak}} \qquad \ldots (13)$$

where, m represents the mass of the jumper, g represents gravitational force = 9. 81m/s$^2$, $v_{t_o}$ refers to take-off velocity and $v_{t_{peak}}$ represents peak velocity. Similarly $h_{t_o}$ and $h_{t_{peak}}$ represents the height at the time of take-off and the height of the jump at peak. Since the vertical velocity is zero at the peak of the jump, Eqn. 13 becomes

$$h_{t_{peak}} - h_{t_o} = \frac{v_{t_o}^2}{2g} \qquad \ldots\ldots (14)$$

Thus the action of the jumper as a projectile is expressed using laws of motion. In this case we consider the take-off velocity as the starting time and final velocity as the ending time. Hence the equation

$$v_f - v_i = -g(t_f - t_i) \qquad \text{..... (15)}$$

Assuming the jumper's centre of mass is same during take-off and landing, and $v_f = -v_i$, this reduces the above equation as

$$v_{t_o} = \frac{g t_d}{2} \qquad \text{..... (16)}$$

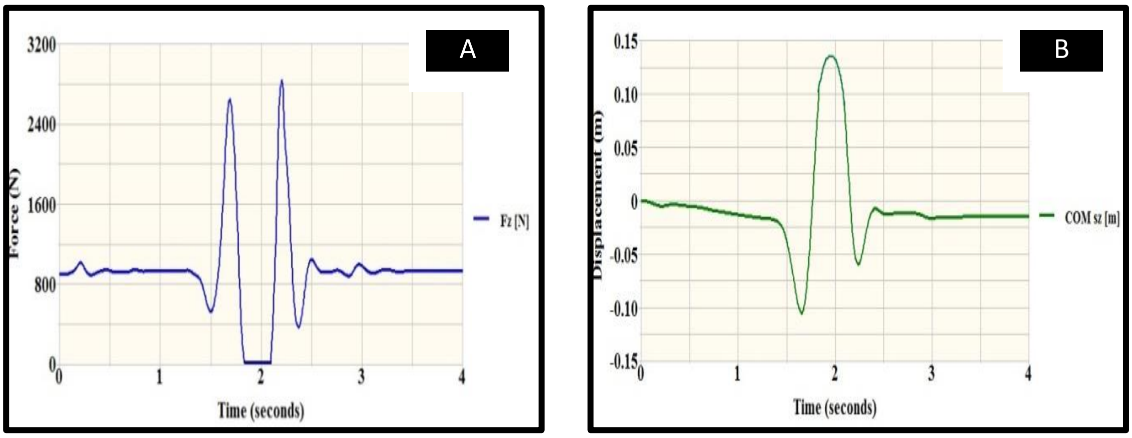


**Figure 20 Variation of A) force and B) displacement with time.**

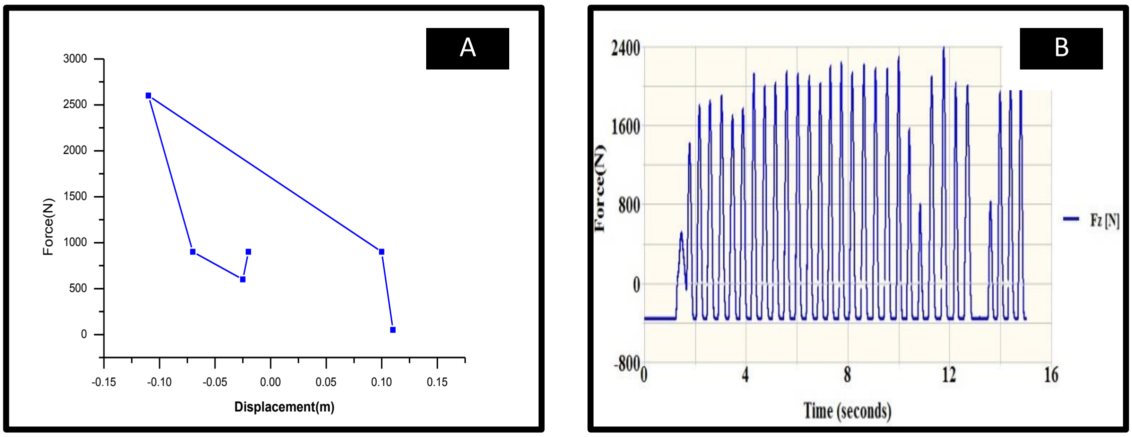


**Figure 21 Variation of A) displacement with time B) Continuous vertical jump**

And the take-off velocity is solved using Eqn. 14. Here $t_d$ represents time difference between take-off and landing.

From Figure 20, the centre of mass displacement with respect to force is realized. In continuous standing vertical jump only a single peak is observed. In this case the force rises gradually beyond mg and the single peak is due to bending of the knees before landing [23]. The curve peaks can be modified by changing the bending motion of the knees during landing. In jumping only a single peak is acquired because at initial heel strike the peak magnitude is attributed to the significant bending of the knees to absorb the impact force for lifting off the ground. The leg does not stretch properly during the stance phase. With respect to that the centre of mass reaches its maximum point causing the ground reaction force to go down resulting in a single peak ground reaction force curve.

## 4.7 Suggestive remedies and design for rehabilitation of a subject suffering from Spastic Quadriplegia

Spastic quadriplegia is a neurological disease which makes the muscles weak and stiff creating difficulty in motion (described in section 3.1). The performance parameters (ground reaction force, centre of pressure, power, torque, coefficient of friction and impulse) of this subject have been discussed in Figure4 – fig 11(B). The gait parameters and forefoot function index is also calculated in table 1, 2, 3 and 4 as subject no. 55 and inferred that the subject gait phases are limited due to uncoordinated muscle action. The gait phases as compared to normal subject are found to exhibit 3 times more energy and time duration. From the analysis the following results are noted

a). Gait cycle duration = 1.34s
b). Right stance phase = 0.852s
c). Right swing phase = 0.384s
d). Left Stance phase = 0.576s
e). Left swing phase = 0.804s
f). Forefoot function index = 5.69s

From the above parameters it is observed that the subject faces problem in the front part. Qualitatively it is observed that the subject’s limb movement is not straight. The subject moves the limb by using abductor muscles and then places the foot on the ground for heel strike and other phases of the gait cycle. The subject is unable to control the lateral deviation of the foot. The subject compensates the foot weakness by moving around for lifting the foot. The main effects observed are:

a). Non-uniform musculo-skeletal forces

b). Limited joint flexibility

c). Tight and stiff muscles restrict the muscles to stretch.

Based upon the gait analysis of the subject having Spastic Quadriplegia the following remedial measures are provided:

a). Irrespective of all other remedies continuous physiotherapy

I. Range of motion exercise

II. Cycling

III. Swimming (making frequent repositioning)

b). Customized insoles made up of polytetrafluoroethylene/ silicon for shock absorption and friction management. The insoles must be gripped to prevent sliding of the foot.

c). Upright supportive structure holding the leg muscles to reduce the shaking nature of the muscle

d). Proper orthopaedic shoes with a certain degree of arc on it in order to improve the function of the foot.

## 4.8 Suggestive remedies and design for rehabilitation of a subject who is orthopedically handicapped with 50%

The subject referred in this case had an accident 8 years ago and has a grafted lower limb structure.

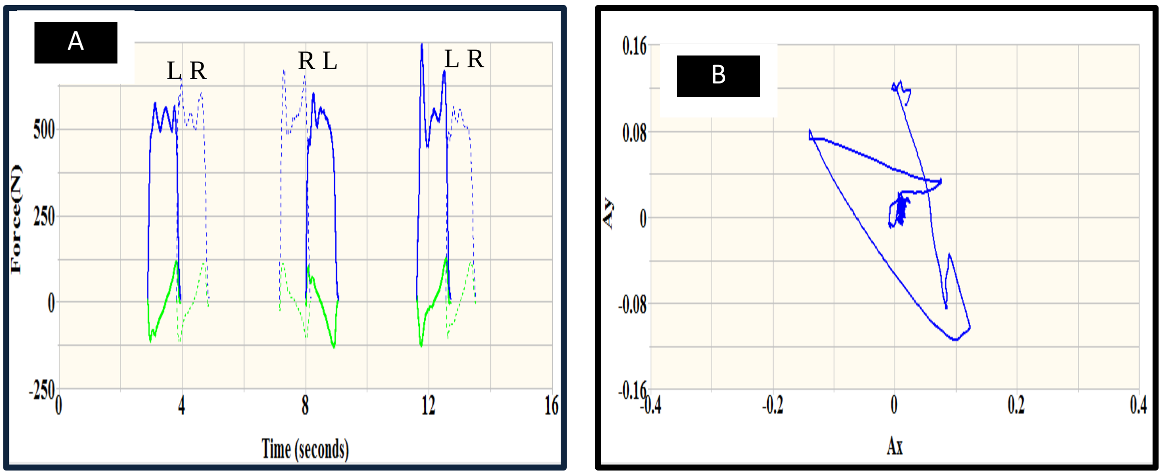


**Figure 22 Variation of A) force with time and B) centre of pressure**

The subject's heel is deformed such that it could not touch the ground. The foot is always in a state of heel rise. According to physiotherapist the ankle joint can be improved by slowly trying to apply stress on the heel as much as possible. The subject is unable to perform dorsiflexion. But there is slight movement on the ankle joint. As the subject is not performing the heel strike phase, the foot slaps on the ground and then lifts the toe from the ground. However, the subject's ground reaction force pattern is observed to be very fluctuating with no sharp humps. The ground reaction force curve shown in Figure 24 does not exhibit similar pattern for both the foot. The first gait cycle represents the ground reaction force pattern of left foot (one with grafting), followed by right foot. The impaired foot reveals only toe-placement and toe-off phase. Due to this the absences of heel strike and mid stance phase, the force magnitude attains maximum during toe-placement and the curve fluctuates during toe-off phase due to dragging the foot on the ground before swinging. The centre of pressure curve in Figure 25 shows left lateral deviation, revealing the chances of sway. The stability is disturbed by straight line of action acting along the left side of the subject.  In case Figure 26 the power dynamics shows that the subject requires high power during braking the motion of the foot. The subject struggles in decelerating the swinging foot motion with longer time duration and accelerates for swinging in a less duration.

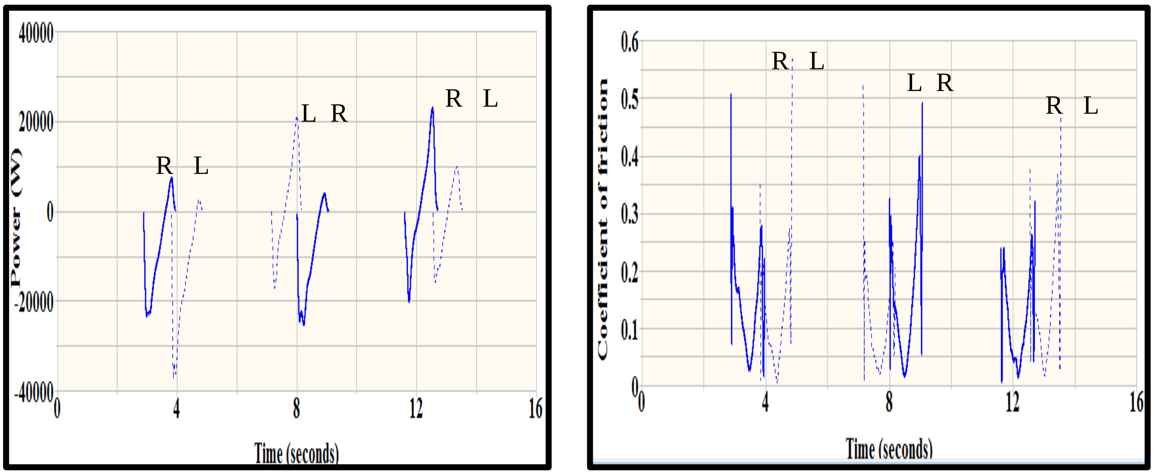


**Figure 23 Variation of A) power with time and B) COF with time**

As the subject comes back after turning around the force platform, the second gait cycle shows same pattern with high swinging power for the normal foot.

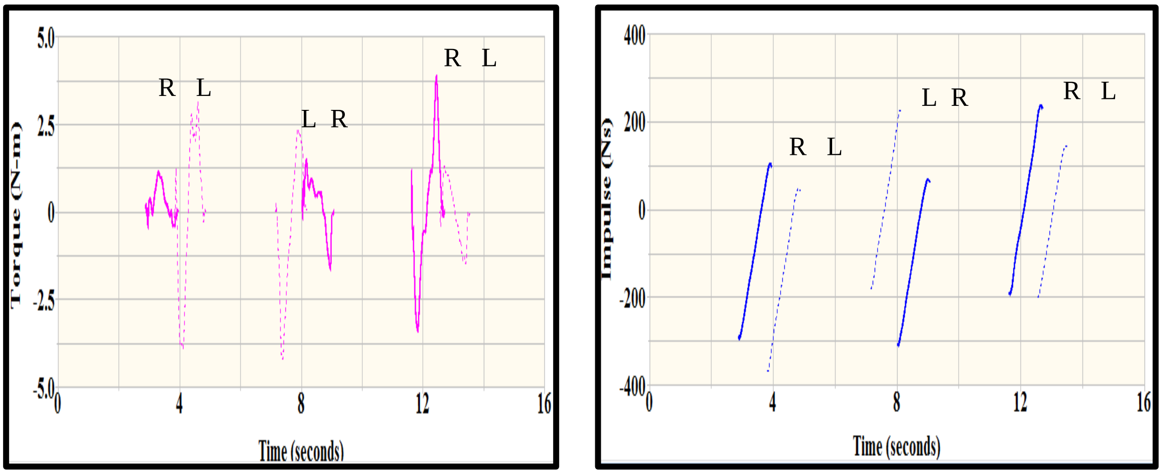


**Figure 24 Variation of A) Torque and B) Impulse with time**

Similar is the case for 3rd gait cycle as well. From Figure 27 it is observed that the subject's impaired foot faces higher coefficient if friction with initial strike due to the plantarflexion of the foot. However for the normal foot the coefficient of friction is highest in the toe-off phase which interacts with the ground to overcome the frictional force.

In Figure 28 the torque is observed to exhibit very high variation for the impaired foot. For normal foot it is not the case. Since no heel strike occurs, so no torque acts on the heel pad and hence the torque acts on the forefoot only. Due to this reason only a single phase of positive clockwise movement is observed in the first part of the first gait cycle. Similarly in second and third gait cycle small phase of torque is observed due to pushing back on the ground in order to acquire energy for lifting off the ground. As the impulse depends upon the amount of force in a certain period of time, the impulse curve shown in Figure 29 shows high impulse.With this information an orthotic design of shoe insert is prepared in Solid works to meet the requirement. The performance gait parameters of this subject are

a). Gait cycle duration = 1.733s
b). Right stance phase = 1.067s
c). Right swing phase = 0.666s
d). Left stance phase = 1.057s
e). Left swing phase = 0.845s
f). Forefoot function index = 6.1s

The orthotic model designed resembles the leaf spring principle. It can be inserted inside the shoes. The design is in such a manner that it possesses high stiffness and high fatigue

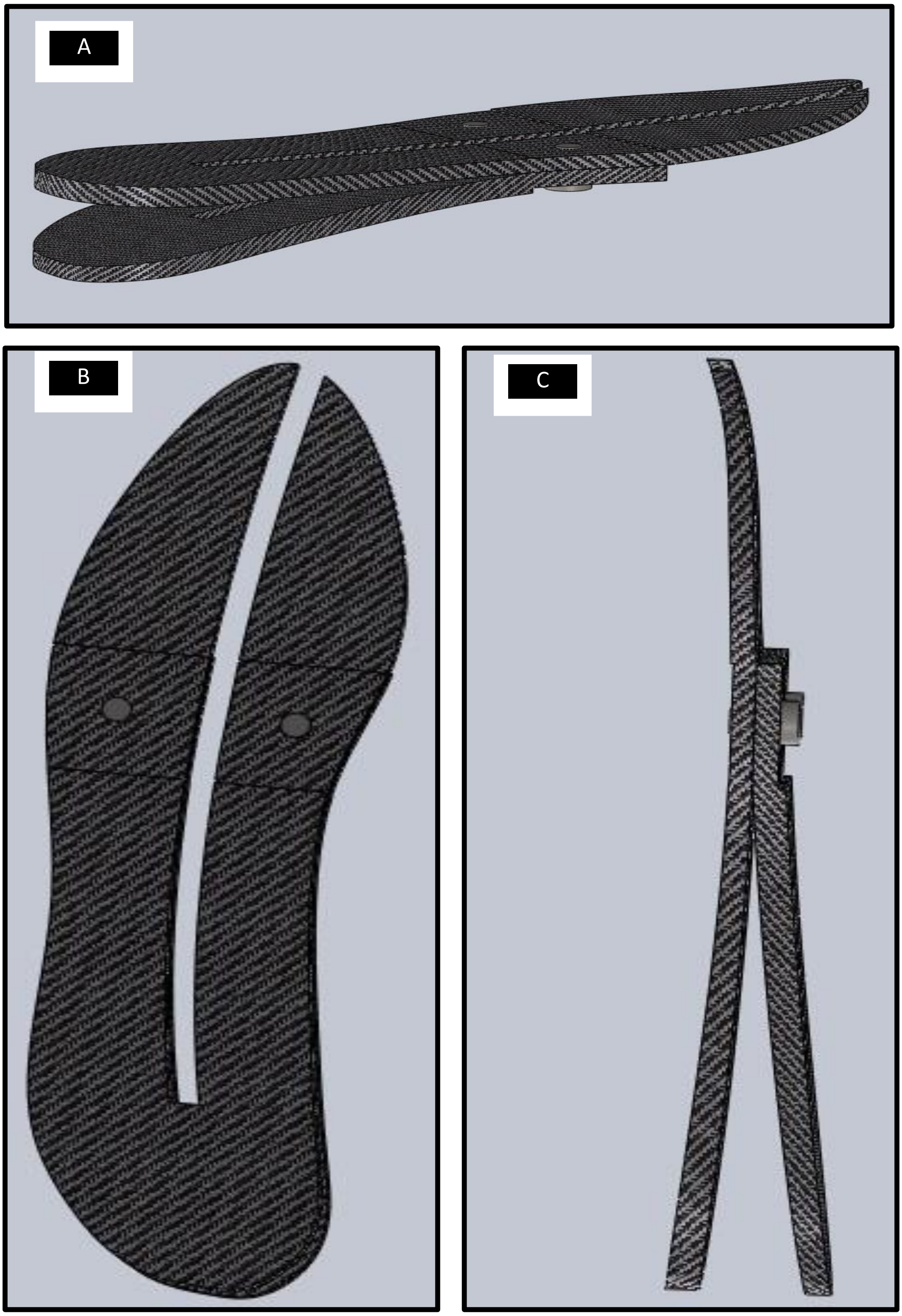


**Figure 25 A) 3D view B) Front view C) Side view of the orthotic design**

life. Apart from these properties the material must be light weight. Hence, materials like titanium alloys/ carbon fibres reinforced light metals can be used to fabricate the orthotic sole. This orthotic can be directly inserted in any shoes. The only thing that must be taken care is the size of the shoe which must be a larger than the actual size of the foot. This kind of orthotic design has been used by many other researchers, but this design eliminates other extra components. Other orthotics requires the assistance of various components such as clamps to hold the orthotics against the foot, rods which align the orthotic base with the leg and external metallic joints around the ankle/ custom made shoe. All these complexities are eliminated by this design which is expected to perform simply as a shoe insert according to preference given by the subject.

# CONCLUSION

The gait analysis of the performance parameters like ground reaction force curve shows the difference between a normal and pathological gait. Any kind of foot pain, injury, sprain etc can be easily visualized from the ground reaction force curve. For normal gait it shows smooth twin peaked curve and fluctuating peak curves in case of pathological gait. This is clearly observed from the analysis. The centre of pressure curve is used to study the postural stability of the subject. It determines the sway nature of the body. The fluctuations at the centre i.e at 0 determine the variation of the body balance of the subject. The coefficient of friction in turn examines the difference between the force applied by a subject and frictional force opposing it. From the analysis it is observed that the frictional force is higher than the normal force applied by the subject. The torque provides thee information regarding the amount of torque applied by the ground. In case of differently abled subject the torque is found to be higher and it causes inversion of the foot. Power dynamics gives an indication of power utilization in order to achieve desired gait activity. The power utilization of differently abled subject is 100 times more than a normal subject. And the impulse applied defines the stress imposed on the specific phases of the gait cycle. The impulse function is determined by forefoot function index which specifies that the impulse is higher in forefoot part for both the case of normal and differently abled subject. From the analysis of gait phases it is noted that the initial planted foot’s stance phase duration is maximum among all other gait phases. A Barefoot walking utilizes larger amount of force compared to walking in shod. When the dynamics of walking is considered it is observed that slower gait pattern results in low stress upon the plantar surface of the foot making it suitable with less amount of energy absorbed.

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